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**MEASUREMENT OF LUNG FUNCTION USING
MAGNETOMETERS. I. PRINCIPLES AND
MATHEMATICAL MODELING.**

D. L. Vawter

W. F. Miner, CAPT, MC, USN

Commanding Officer

Naval Medical Research Institute

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Acknowledgments

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Dr. Vawter served as an ASEE Summer Faculty Research Fellow in the Hyperbaric Medicine Program Center during 1979. His permanent address is School of Engineering Science and Mechanics, Georgia Institute of Technology, 225 North Ave., N.W., Atlanta, GA 30332.

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REPORT DOCUMENTATION PAGE		READ INSTRUCTIONS BEFORE COMPLETING FORM
1. REPORT NUMBER (14) NMRI-79-78	2. GOVT ACCESSION NO. AD-A083 910	3. RECIPIENT'S CATALOG NUMBER
4. TITLE (and Subtitle) MEASUREMENT OF LUNG FUNCTION USING MAGNETOMETERS I. PRINCIPLES AND MATHEMATICAL MODELING		5. TYPE OF REPORT & PERIOD COVERED MEDICAL RESEARCH PROGRESS REPORT
7. AUTHOR(s) (12) Donald L. Vawter		8. CONTRACT OR GRANT NUMBER(s) (12) 551
9. PERFORMING ORGANIZATION NAME AND ADDRESS Naval Medical Research Institute Bethesda, Maryland 20014		10. PROGRAM ELEMENT, PROJECT, TASK AREA & WORK UNIT NUMBERS (17) M0099/PN, 002/8011 Report No. 2
11. CONTROLLING OFFICE NAME AND ADDRESS Naval Medical Research & Development Command Bethesda, Maryland 20014		12. REPORT DATE (11) November 1979
14. MONITORING AGENCY NAME & ADDRESS (if different from Controlling Office) Bureau of Medicine & Surgery Department of the Navy Washington, D.C. 20372		13. NUMBER OF PAGES 48
		15. SECURITY CLASS. (of this report) UNCLASSIFIED
16. DISTRIBUTION STATEMENT (of this Report) APPROVED FOR PUBLIC RELEASE AND SALE: DISTRIBUTION IS UNLIMITED. (13) UNCLASSIFIED		
17. DISTRIBUTION STATEMENT (of the abstract entered in Block 20, if different from Report)		
18. SUPPLEMENTARY NOTES		
19. KEY WORDS (Continue on reverse side if necessary and identify by block number) magnetometers, lung function, respiration, mathematical modeling, hysteresis, ventilation		
20. ABSTRACT (Continue on reverse side if necessary and identify by block number) The use of magnetometer systems to predict pulmonary air volumes was studied. The sensitivity of the system to angular rotations was found to be small because magnetometers are normally mounted to the subject. Seven mathematical models for predicting volume were investigated, with the finding that two-parameter models are adequate for quiet breathing. For complicated maneuvers, a three-parameter model is necessary, and for predictive usefulness, the model must be calibrated using a complex maneuver.		

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Results of studies of the dimensional changes, and the correlation between the dimensional changes, that occur during normal breathing are presented.

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MEASUREMENT OF LUNG FUNCTION USING MAGNETOMETERS.

I. PRINCIPLES AND MATHEMATICAL MODELING

Donald L. Vawter, Ph.D.

Previous studies by Konno and Mead (1), and Robertson et al. (2) have shown the feasibility of using magnetometers to sense dimensional changes occurring during respiration. Robertson et al. discussed certain mathematical functions which predict lung volume from the observed dimensional changes of the thorax and abdomen. There remain, however, many unanswered questions on the use of magnetometers to calculate volume. In this study we have attempted to answer some of the questions and in the process have raised some new ones. In particular, we shall assess the effects of magnetometer rotation, discuss alternative models, describe a new method of determining the unknown constants in the models, and discuss the interdependence of the various respiratory movements and its influence on the choice of model parameters.

SENSITIVITY OF THE MAGNETOMETER SYSTEM TO ANGULAR ROTATION

In general, during the use of the magnetometer system, it is impossible to avoid some relative angular rotation between magnetometer pairs. If the magnetometers are to measure length changes, one would hope that the system is insensitive to these angular changes. We have measured the change in output of the magnetometer system as the angular orientation of the transmitter and receiver was changed. There are, of course, three types of angular rotation. We refer to these as rotations about the X, Y, and Z axes (see Fig. 1 for definitions of the axes). First, consider rotation about the X axis (i.e. the axis joining the centers of the magnetometer pair). Rotation about this axis causes tremendous changes in output and must be avoided at all costs. Fortunately, the magnetometers as conventionally mounted should have little tendency to rotate about this axis. Of more interest is rotation about the Y and Z axes. We found that rotation about the Y and Z axes were equivalent, and rotation about the Y axis will not be discussed.

To measure angular effects, we mounted a pair of magnetometers a fixed distance apart (20, 25, or 30 cm) and rotated either the receiver or transmitter. There was, indeed, a change in output with angular change, as can be seen in Fig. 2. Because of limitations in the experimental apparatus, it was not possible to achieve pure rotation; in fact, the magnetometers moved closer together as the angle was increased. If we take this effect into account, we find the results shown in Fig. 3. The solid curve is the output expected if rotation has no effect. In other words, the solid curve is reflecting the decrease in the separation between the transmitter and receiver as the angle increases. Deviation from the solid line is then the effect of rotation. Notice that as long as the rotation is less than 35° , the maximum error in the output is 15 mV, which corresponds to 0.05 cm. It is unlikely that relative angular changes during respiration would exceed these values.

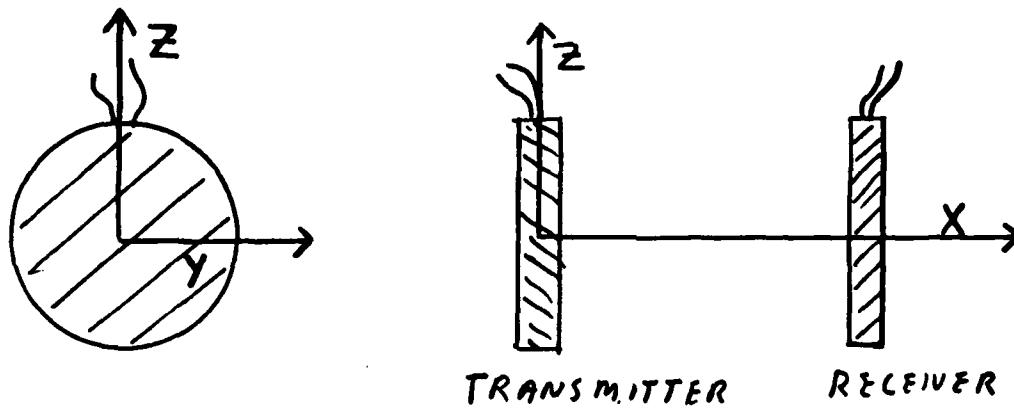


Fig. 1. Definition of axes for angular rotation study.

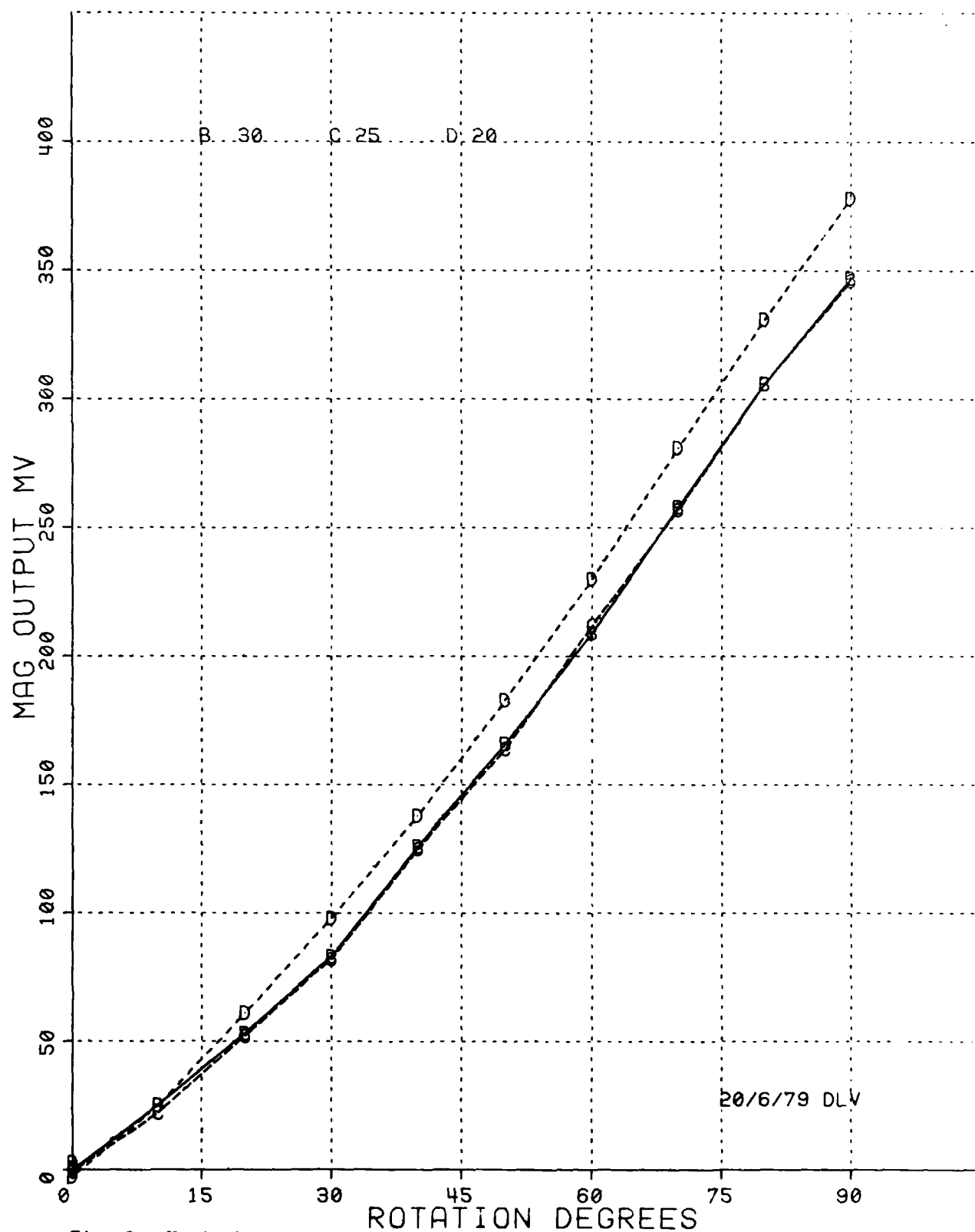


Fig. 2. Variation in magnetometer output versus rotation of receiver. Separation between transmitter and receiver varies and is given by "B" 30 cm, "C" 25 cm, and "D" 20 cm.

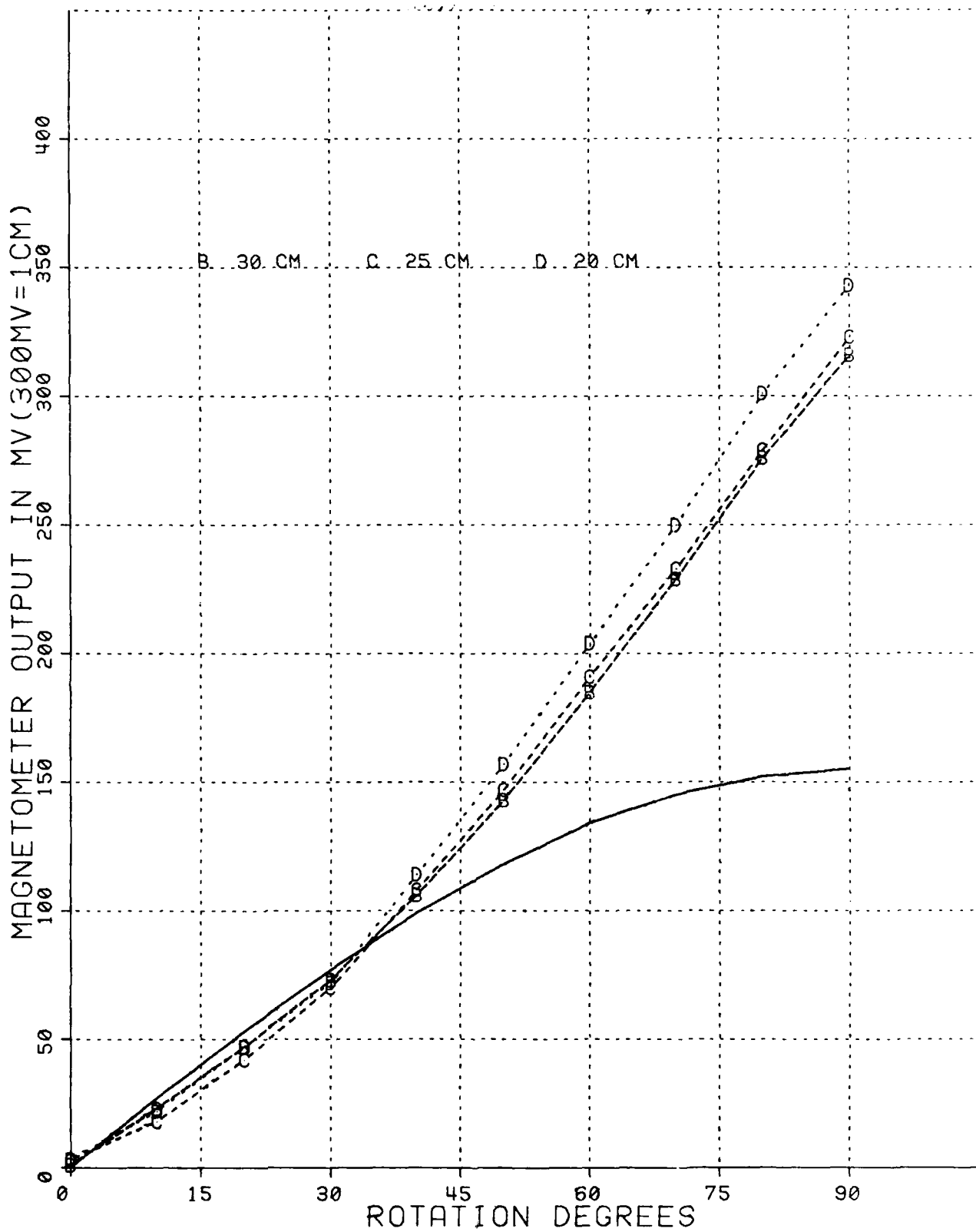


Fig. 3. Correction of magnetometer output for changes in separation during rotation. Solid curve is the output to be expected if angular rotation has no effect.

It is, however, quite likely that the faces of the transmitter and receiver will not be parallel when attached to the body, and it is possible that the linearity and/or gain of the system may be adversely affected. To test this, we rotated the receiver (or transmitter, it made no difference) to a specified angle and then varied the separation between the transmitter and receiver. The results are shown in Figs. 4 and 5. In Fig. 4, equal gain is implied because all lines are parallel. Also, the essentially straight curves show linearity is not adversely affected by rotation. Figure 5 plots a small range of the results on an expanded scale showing results similar to Fig. 4.

We conclude from the above study that angular rotation is unlikely to cause significant errors as long as rotation about the X axis is avoided. Also, if one had magnetometer separations much less than 20 cm, additional testing would be necessary to determine angular effects.

USE OF MAGNETOMETERS TO CALCULATE VOLUME

The capability to predict volume changes during respiration using the four pair magnetometer system would be useful. Robertson et al. (2) have described this concept in detail. To use the magnetometer data to predict volume changes, one can use at least two approaches.

One may model the anatomy of the chest and abdomen in some particularly simple way and then find the unknown constants in the anatomic model by minimizing the difference between the observed spirometer volumes and those calculated with the anatomic model. Robertson et al. have used this approach and have modeled the chest and abdomen as elliptic cylinders. They have chosen the following equation to predict the volume:

$$V = K_1 * DC - K_2 * DC / DA - K_3,$$

where DC is the cross-sectional area of the chest; DA is the cross-sectional area of the abdomen; V is the volume; K_1 , K_2 , and K_3 are unknown anatomical parameters. K_1 corresponds to the distance between the symphysis pubis and the sternal notch, K_2 corresponds to the volume of the abdomen, and K_3 corresponds to the non-gaseous volume of material in the chest. The constants K_1 , K_2 , and K_3 are determined by minimizing the squared error between the observed spirometer volumes and those predicted using the equation. The advantage of this anatomical model is that the constants have physical meaning; therefore one hopes the model has predictive value. Robertson et al. have used a non-linear iterative scheme to calculate the three constants. It will be shown shortly that such an approach is an unnecessary complication.

Another approach is to attempt to calculate volume without regard to an anatomical model. This is less satisfying, but it may be possible to obtain better curve fits using such an approach, because the form of the model equation is less restrictive. However, whether such an approach has any predictive value must be determined. It might be, for example, that the assumed curve may accurately model the curve from which the constants were determined, but fail completely in predicting the volume of another breath. To test the validity of non-anatomic modeling,

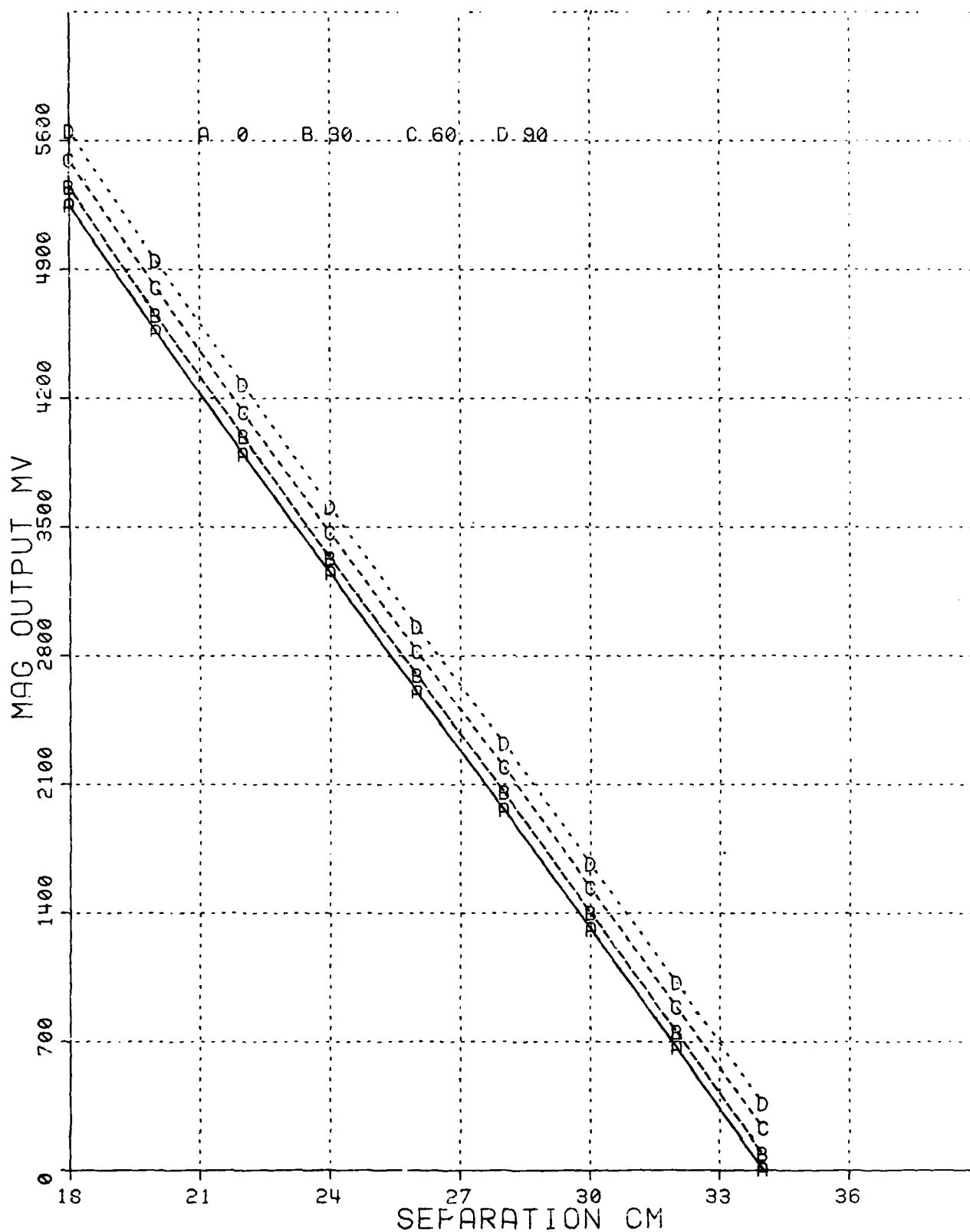


Fig. 4. Linearity of output for various angular orientations. Angular rotation for curves are "A" 0 degrees, "B" 30 degrees, "C" 60 degrees, and "D" 90 degrees. Note curves are parallel.

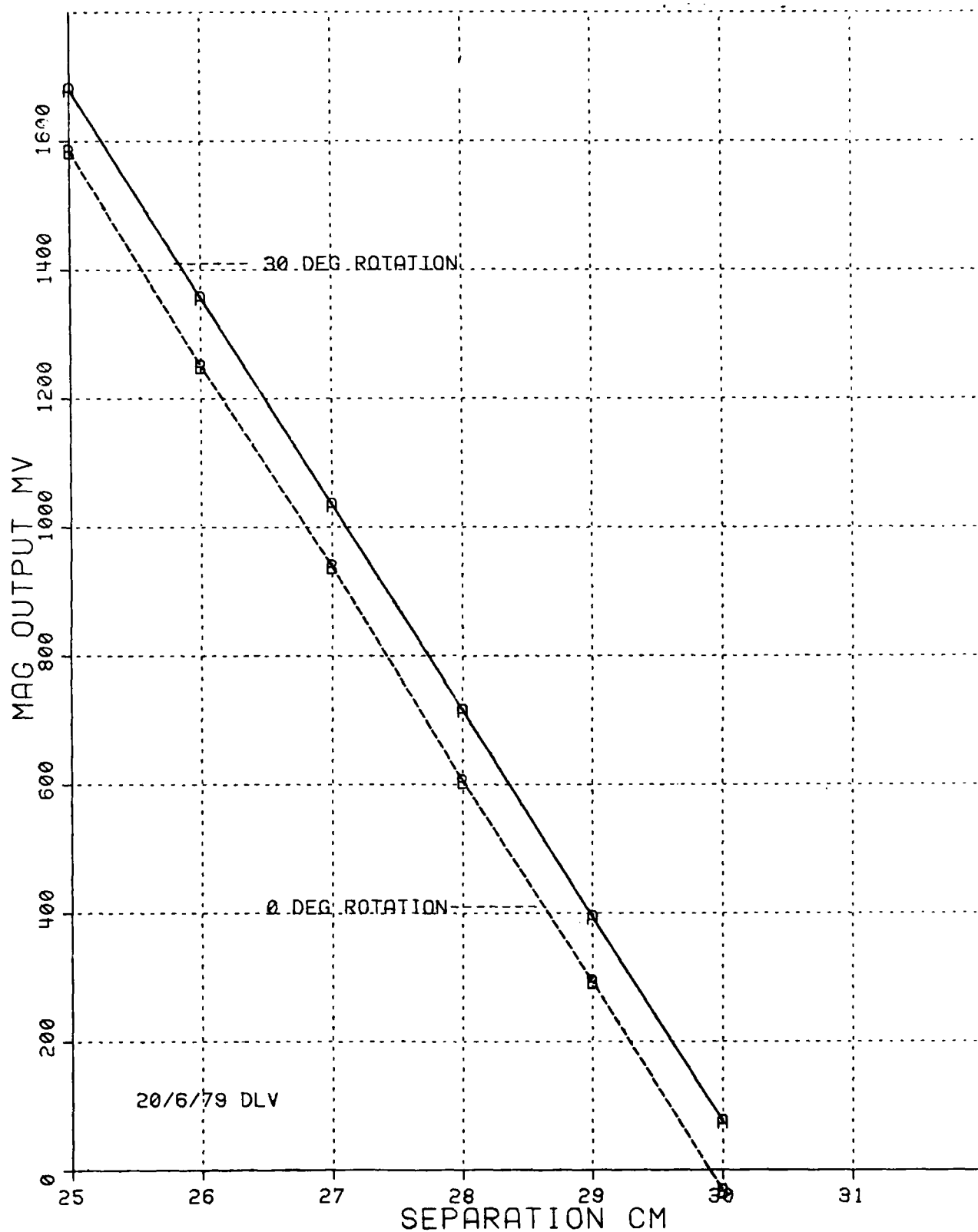


Fig. 5. Expanded scale showing a portion of Fig. 4. Notice neither linearity nor gain is affected by angular rotation. The intercept, however, does change.

we hypothesized and tested seven models. The equations for the models are given below:

$$\text{Model 1: } V = K_1*(DC - DCO) - K_2*(DA - DAO)$$

$$\text{Model 2: } V = K_1*(APC - APCO) - K_2*(APA - APAO)$$

$$\text{Model 3: } V = K_1 - K_2*(APA - APAO)$$

$$\text{Model 4: } V = K_1*(APC - APCO) - K_2$$

$$\text{Model 5: } V = K_1*DC - K_2*(DC/DA) - K_3$$

$$\text{Model 6: } V = K_1*(DC - DCO) - K_2*(DC/DA - DCO/DAO) - K_3$$

$$\text{Model 7: } V = K_1*DC - K_2*DA - K_3.$$

In these equations, V is volume, DC is the cross-sectional area of the chest, DA is the cross-sectional area of the abdomen, APC is the anterior-posterior (A-P) diameter of the chest, and APA is the A-P diameter of the abdomen. The corresponding values at FRC (measured with callipers) are followed with zero (0). In the above equations, although they are not anatomical models, we still assume the cross-sections of the abdomen and chest are elliptical when calculating areas.

A few comments about the various models are perhaps in order. In Models 1, 2, and 3 the volume is forced to be zero at FRC. Model 5 is the same as the anatomic model of Robertson et al. Models 5 and 6 should be equivalent, differing only in the value for K_3 . Model 7 is a generalization of Model 1, which does not force the volume to go through zero at FRC.

DETERMINATION OF THE UNKNOWN CONSTANTS USING A LEAST SQUARED ERROR CRITERIA

In the previous equations, the undetermined constants K_1 , K_2 , and K_3 must be given numerical values before the equations can be used. The most popular criterion for determining the constants is that the square of the difference between the volume predicted by the model equation and that observed with the spirometer, summed over all sample points, should be a minimum. In other words, we want ESQ to be a minimum where:

$$\text{ESQ} = \sum_{i=1}^N E_i^2$$

with $E_i = V_i - VC_i$.

V_i is the spirometer volume at sample point I and VC_i is the volume at sample point I obtained from the equations of the model. For example, in Model 5 we have

$$VC_i = K_1*DC_i - K_2*(DC_i/DA_i) - K_3.$$

There are several options for determining the constants that will make ESQ a minimum. Robertson et al. have used an iterative scheme based on numerical minimization of ESQ. A simpler method, pursued here, is to minimize the error analytically. Since the constants are related to each other linearly, the resultant system of algebraic equations is

linear. To minimize the error analytically, we must take the partial derivative of ESQ with respect to each unknown constant and set it equal to zero. This results in as many linear equations as unknown parameters. For a three-parameter model the system of equations has the following form:

$$\begin{aligned} S(1)*K_1 - S(2)*K_2 - S(3)*K_3 &= S(4) \\ S(2)*K_1 - S(6)*K_2 - S(7)*K_3 &= S(8) \\ S(3)*K_1 - S(7)*K_2 - S(11)*K_3 &= S(12). \end{aligned}$$

where:

$$\begin{aligned} S(1) &= \text{sum over all } I \text{ of } X(I)*X(I) \\ S(2) &= \text{" " " " } X(I)*Y(I) \\ S(3) &= \text{" " " " } X(I)*Z(I) \\ S(4) &= \text{" " " " } X(I)*V(I) \\ S(6) &= \text{" " " " } Y(I)*Y(I) \\ S(7) &= \text{" " " " } Y(I)*Z(I) \\ S(8) &= \text{" " " " } Y(I)*V(I) \\ S(11) &= \text{" " " " } Z(I)*Z(I) \\ S(12) &= \text{" " " " } Z(I)*V(I). \end{aligned}$$

$X(I)$ is the variable multiplying K_1 in the model equation,
 $Y(I)$ is the variable multiplying K_2 in the model equation, and
 $Z(I)$ is the variable multiplying K_3 in the model equation.
 For the present study, $Z(I)$ is always equal to 1. To clarify by example: In Model 5, $X(I) = DC_i$, $Y(I) = DC_i/DA_i$, $Z(I) = 1$.

The equations generated above can easily be solved for the unknown constants, and in the present study they were solved using Cramer's rule of determinants. It should be strongly emphasized that because the changes in the chest and abdomen measurements are small compared to the original diameters, the equations are nearly singular and at a very minimum, double-precision arithmetic must be used. Also, because of this ill conditioning, although the combination of the constants determined may be quite useful, the values of an individual constant cannot be reliably determined. In other words, the equations can be used to calculate volumes but we cannot compare values of K_2 , for example, between runs, and make any conclusions (e.g. because K_2 is higher for person "A" than for person "B", person "B" must use abdominal breathing and person "A" must use intercostal breathing). With these precautions, one may use this method to determine the constants. The method is much simpler than that used by Robertson et al. because it requires no initial estimates of the constants nor any iterations. It is also preferable because iterative solutions often get "trapped" into solving for the local rather than global minimums if the initial estimates

are not close to the true values. In the present approach, dependence on the skill of the user to make estimates is unnecessary and, hence, the programs can be run without technical knowledge of the modeling process. We next discuss the results obtained using the seven models described.

NONPREDICTIVE USE OF THE MODELS

Before we can assess the predictive value of any of the models, we must determine their adequacy to calculate volume on the breath from which the constants were obtained. Robertson et al. refer to this as calibration studies. Rather than present a large number of figures for a number of subjects, we will discuss one subject in detail. We shall consider two types of respiratory maneuvers in this section: tidal breathing of about 700 to 1000 ml, and a quiet breath followed by a Valsalva maneuver and another quiet breath.

The curve fits for the seven models shown in Figs. 6 through 12 are for the large tidal volume breaths. Notice that all models are adequate for this purpose. The RMS errors for the models are tabulated in Table 1. We found that Models 5 and 6 are indeed equivalent, a fact which assures us that even though the system of equations determining the constants is ill-conditioned, the resultant combination of constants is adequately determined. Model 6 will not be further discussed.

In Figs. 13 through 18, the curve fits of the models determined using the Valsalva data are presented. Notice that the two parameter models (Models 1 through 4) are incapable of fitting the data adequately, showing only qualitative agreement. Models 5 and 7 again show an excellent fit. The details of the curve fits are found in Table 1.

From these figures, we conclude that the two parameter models studied may be adequate for monitoring quiet breathing but are not useful in complicated maneuvers that presumably cause geometrical distortions to the lung and chest, which are not reflected in the models. The three parameter models (5 through 7), on the other hand, do quite well in both maneuvers.

PREDICTIVE VALUE OF THE MODELS

If the models are to be of any quantitative usefulness, they should be able to predict volumes for breaths other than the one from which the constants were obtained. Robertson et al. have tested their model using constants obtained from large tidal volumes (obtained by a rebreathing method) on vital capacity maneuvers and on quiet breathing maneuvers. Because our three parameter Model 5 is equivalent to theirs, we need not repeat their study on predictive values. Model 7 appears to do no better or worse on their data, so we will concern ourselves only with testing the models on complicated respiratory maneuvers, particularly the Valsalva maneuver.

In using the models for prediction, it is quite often necessary to add a constant volume of gas to the model prediction. This is so because we have never forced the end expiratory volume of the spirometer

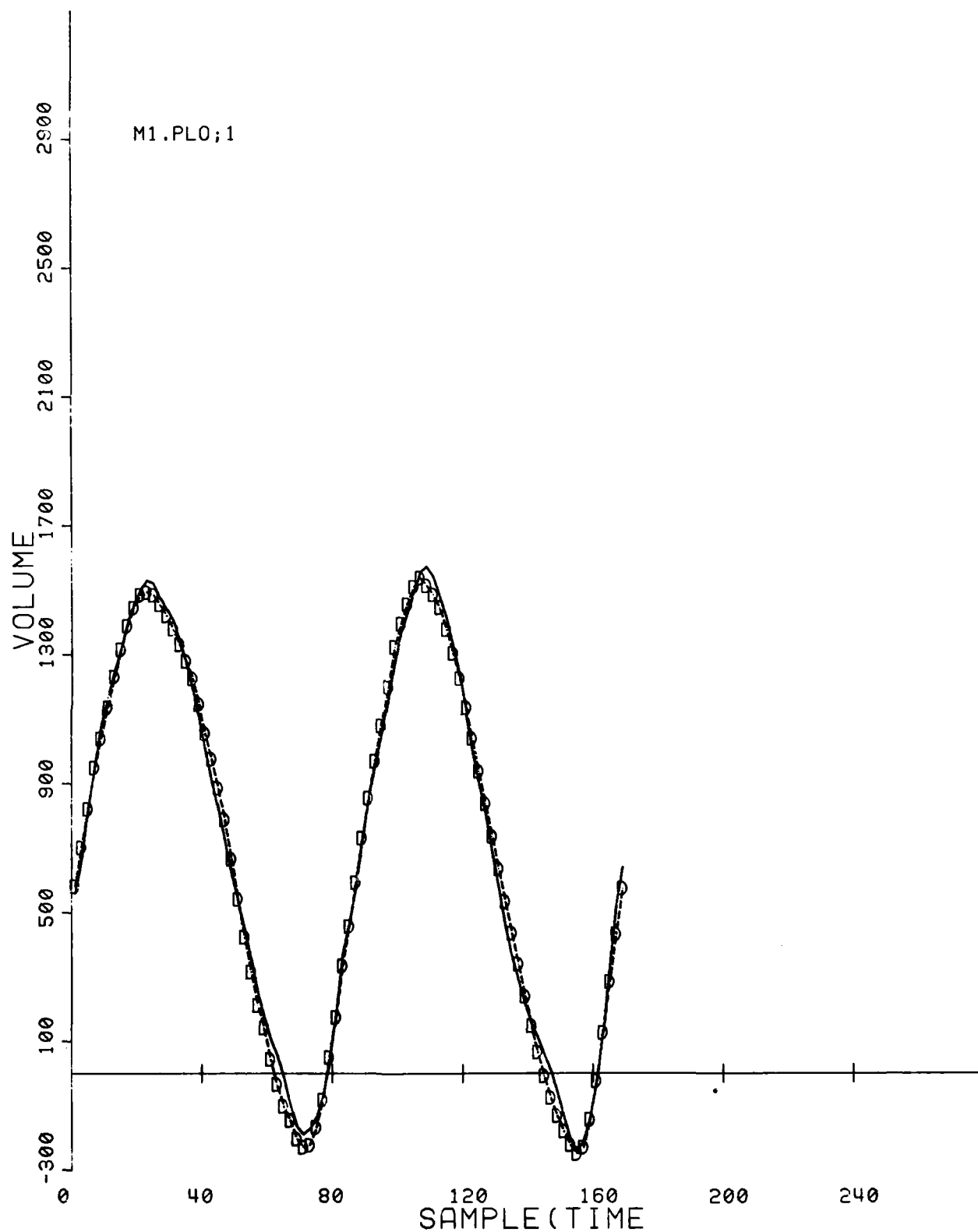


Fig. 6. Tidal breathing. Curve fit for Model 1. Solid curve is the model prediction. Curve "D" is the spirometer output.

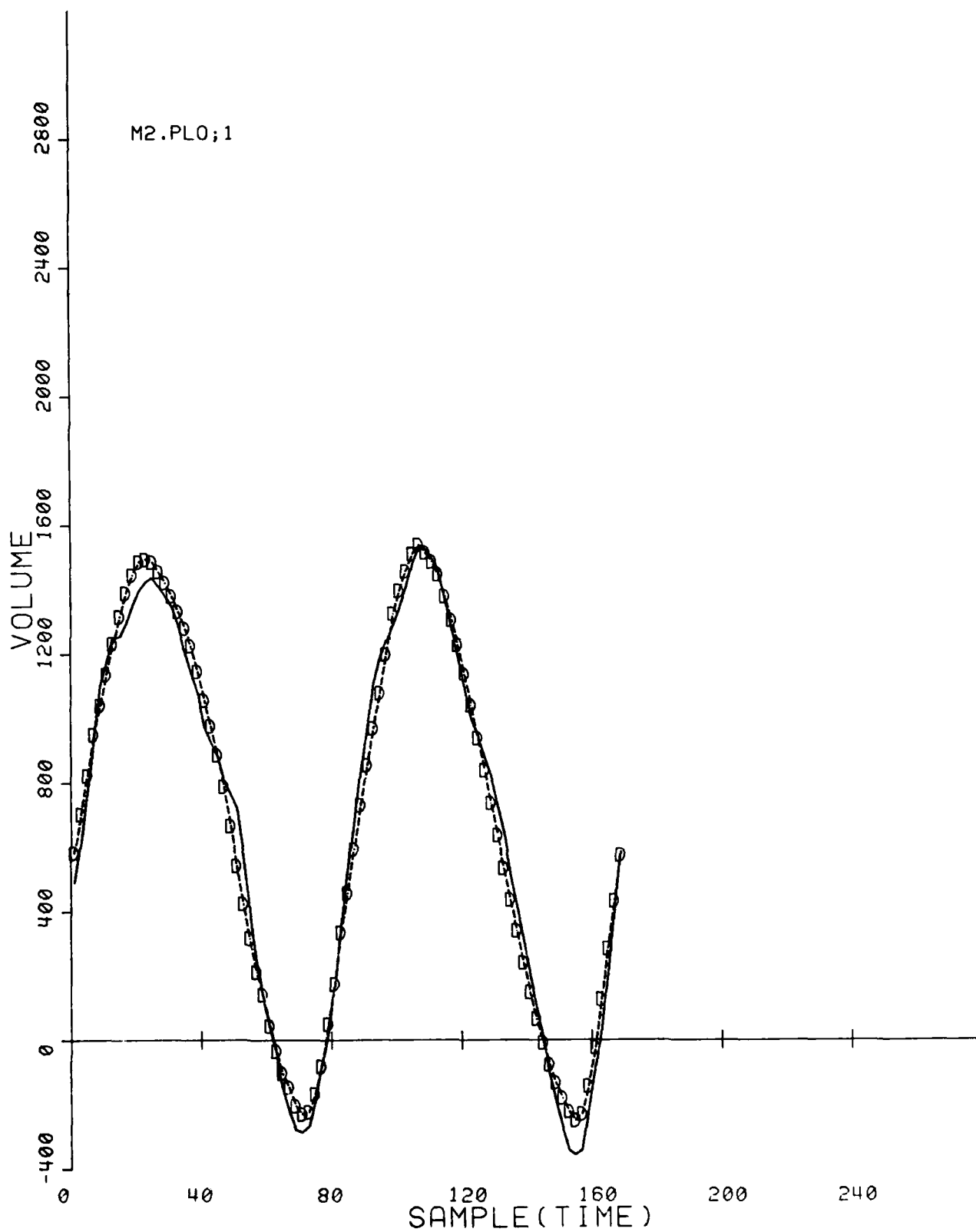


Fig. 7. Tidal breathing. Curve fit for Model 2.

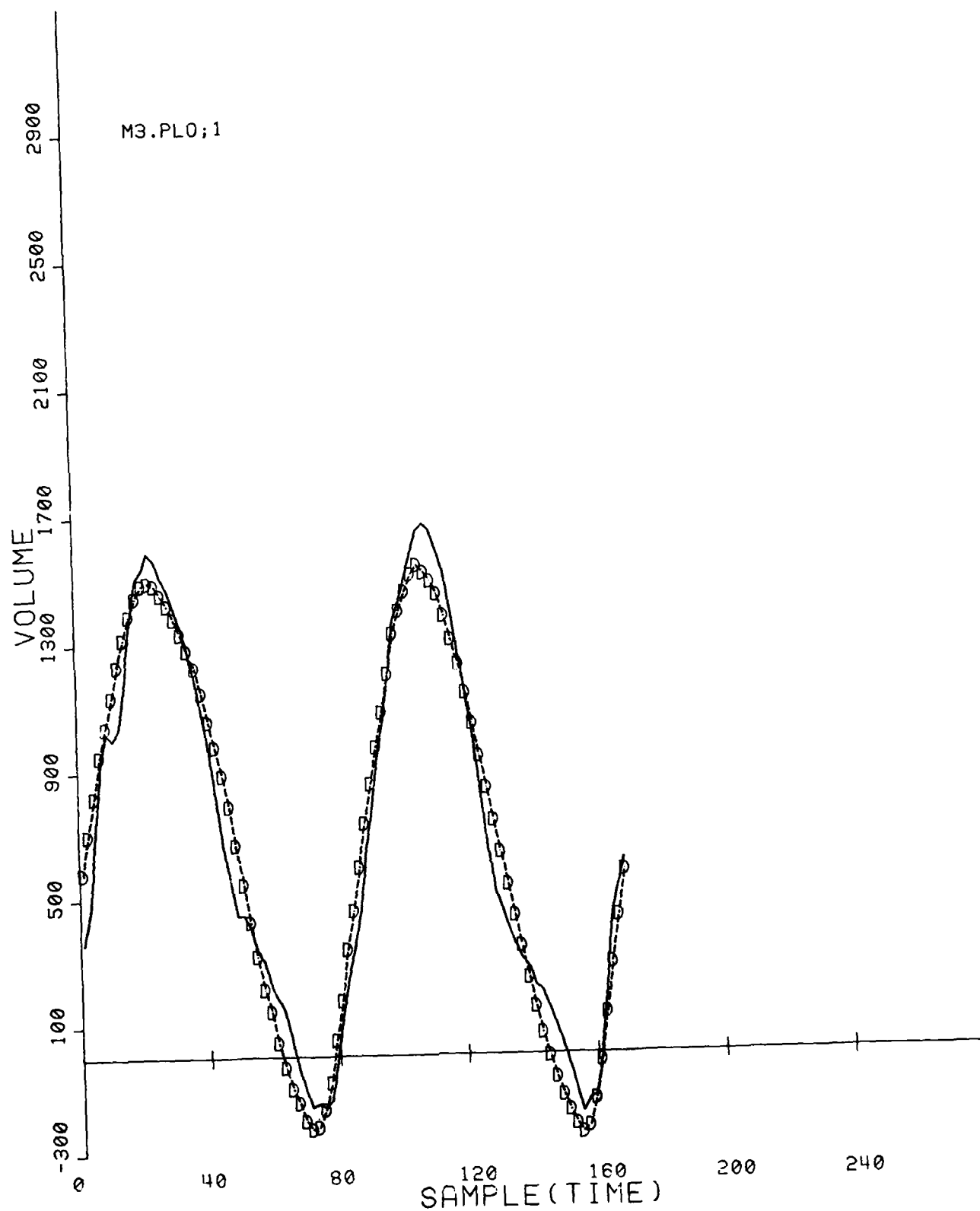


Fig. 8. Tidal breathing. Curve fit for Model 3.

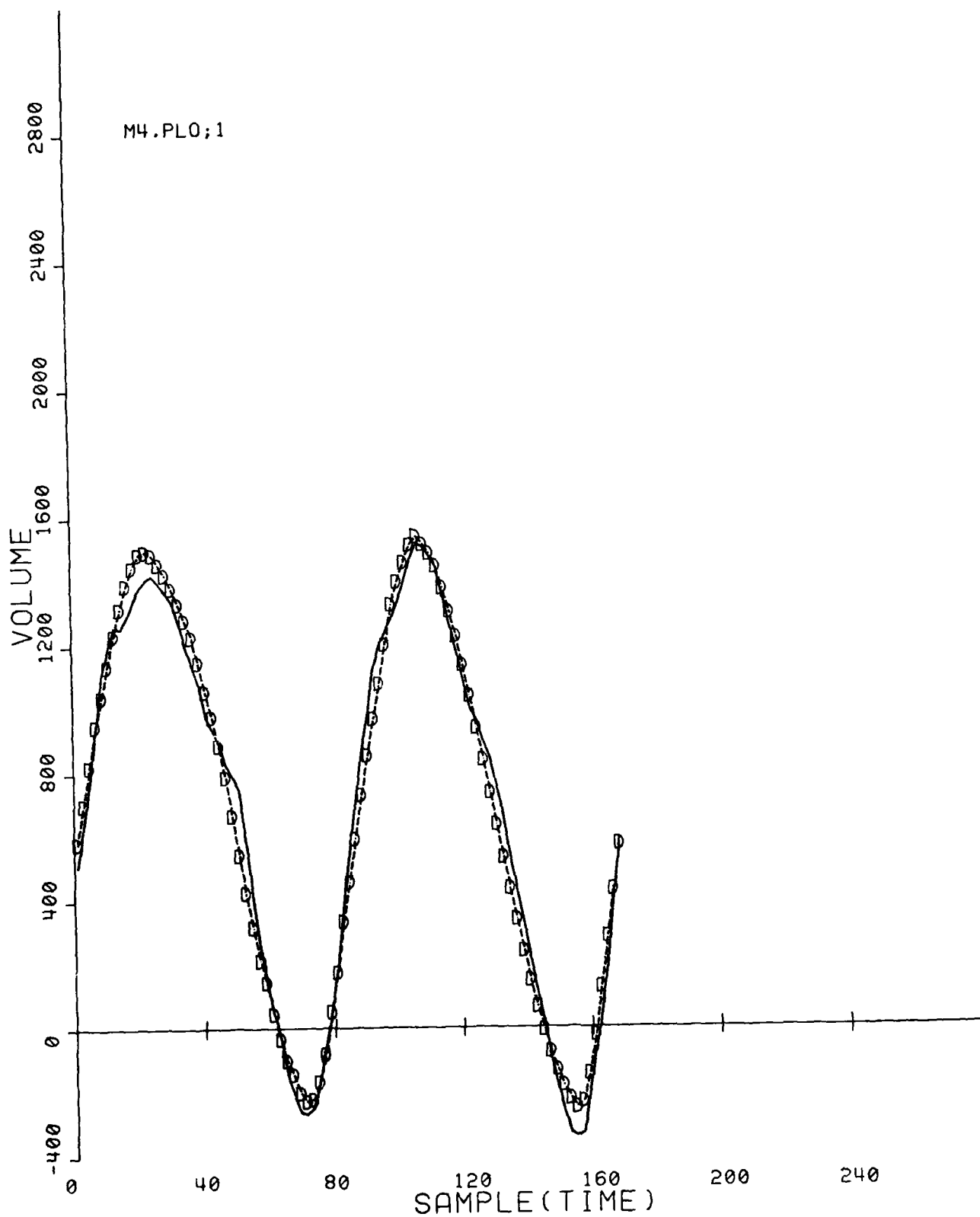


Fig. 9. Tidal breathing. Curve fit for Model 4.

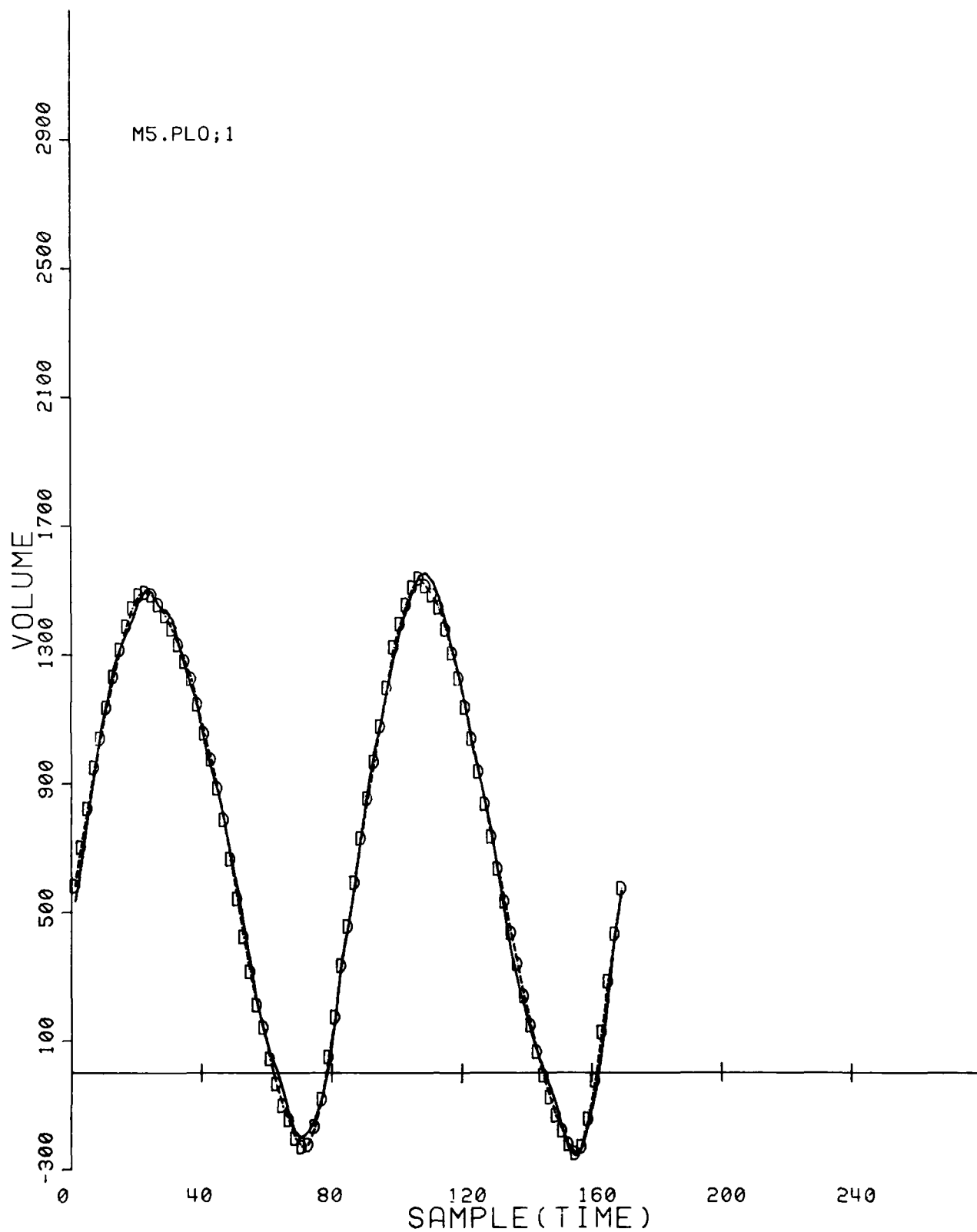


Fig. 10. Tidal breathing. Curve fit for Model 5. This is the same model as used by Robertson et al. (ref. 2).

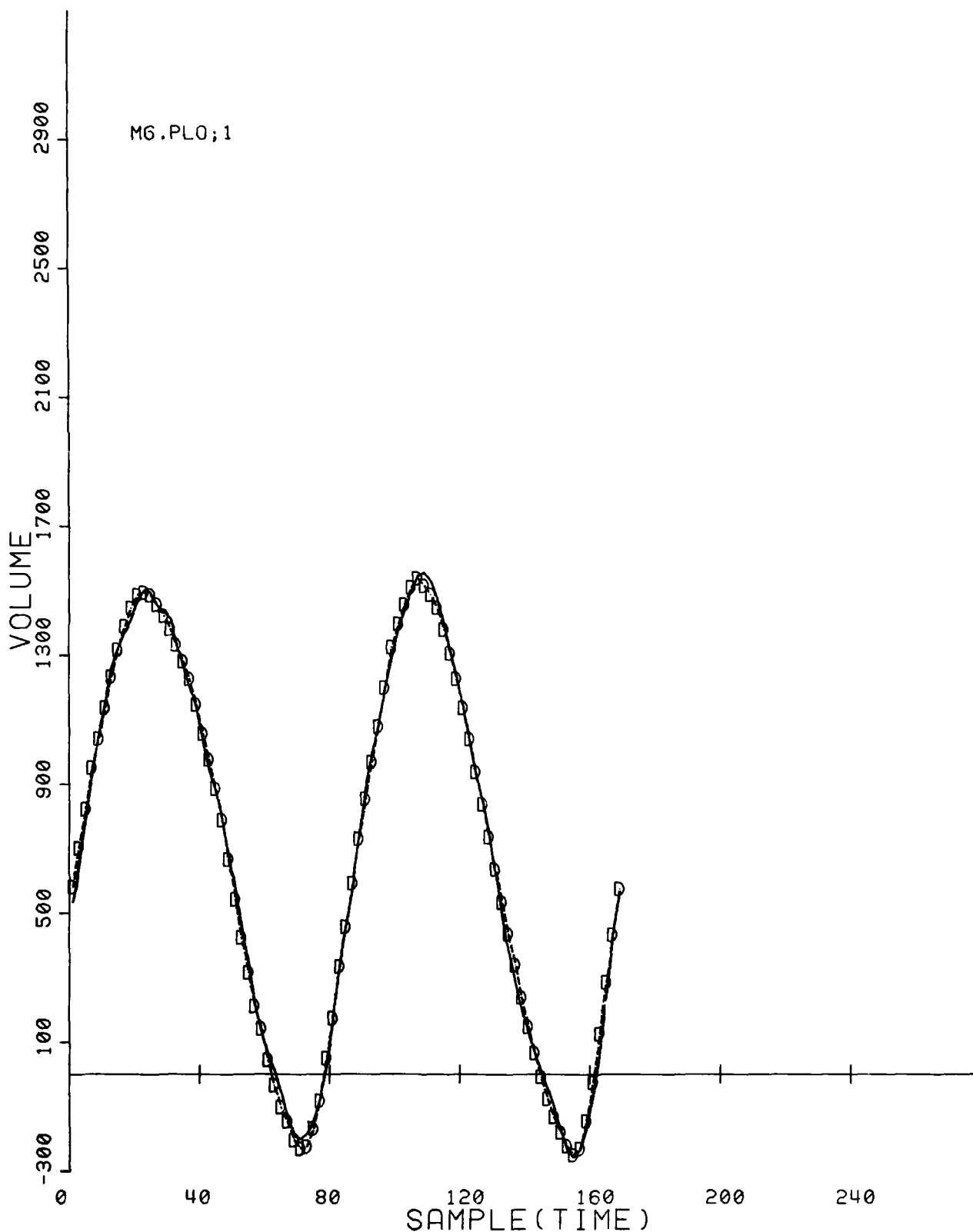


Fig. 11. Tidal breathing. Curve fit for Model 6. Notice Models 5 and 6 are equivalent.

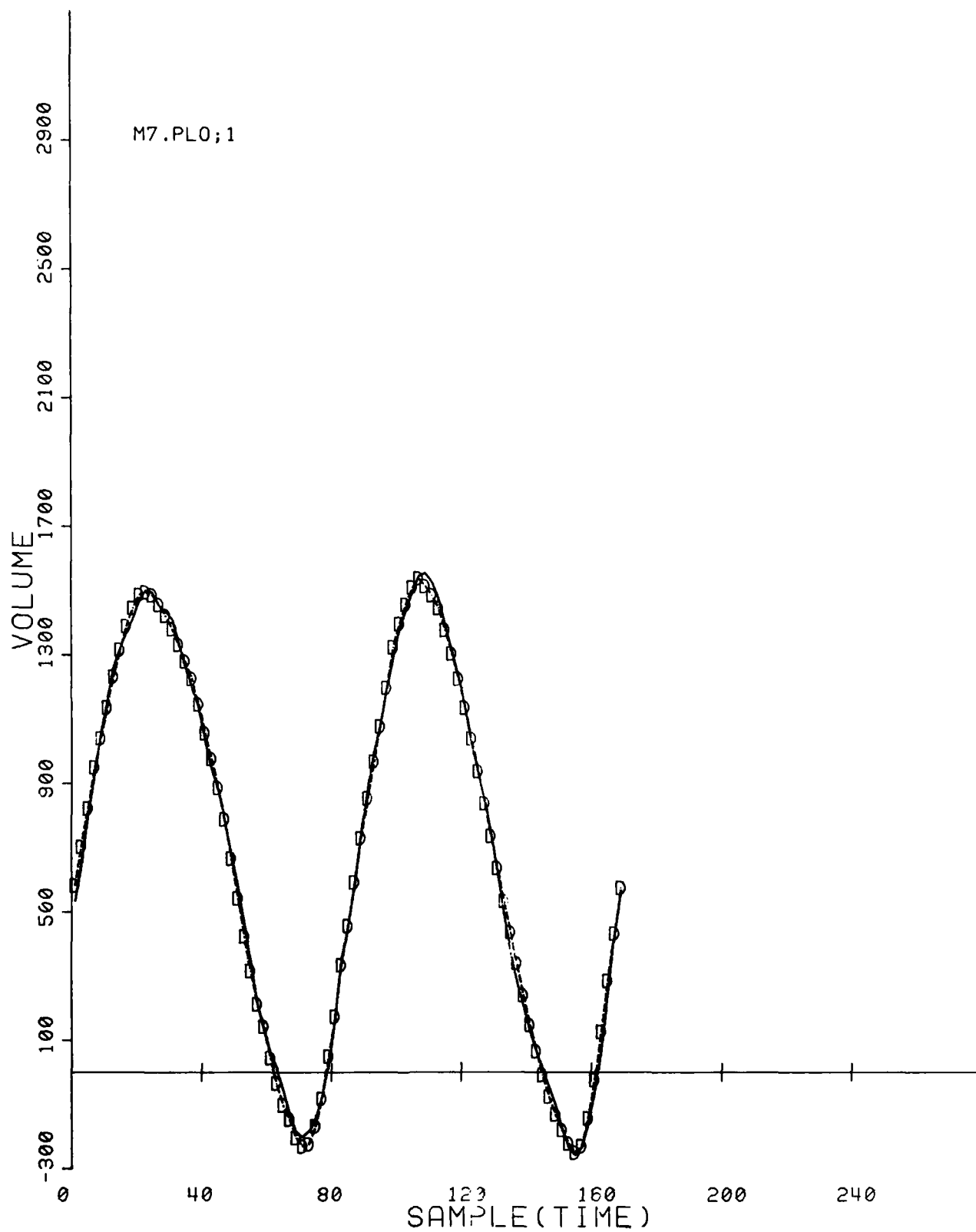
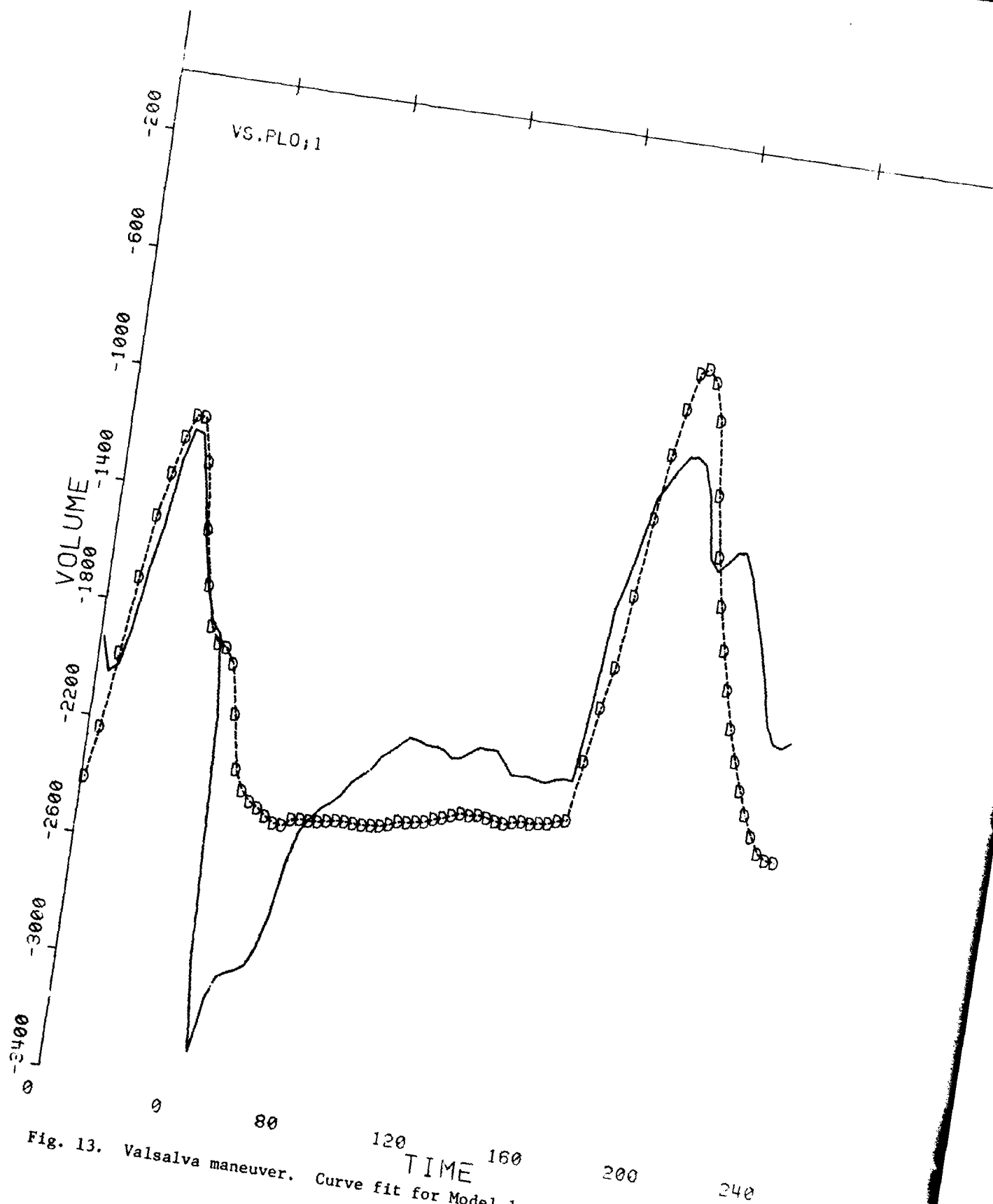


Fig. 12. Tidal breathing. Curve fit for Model 7.



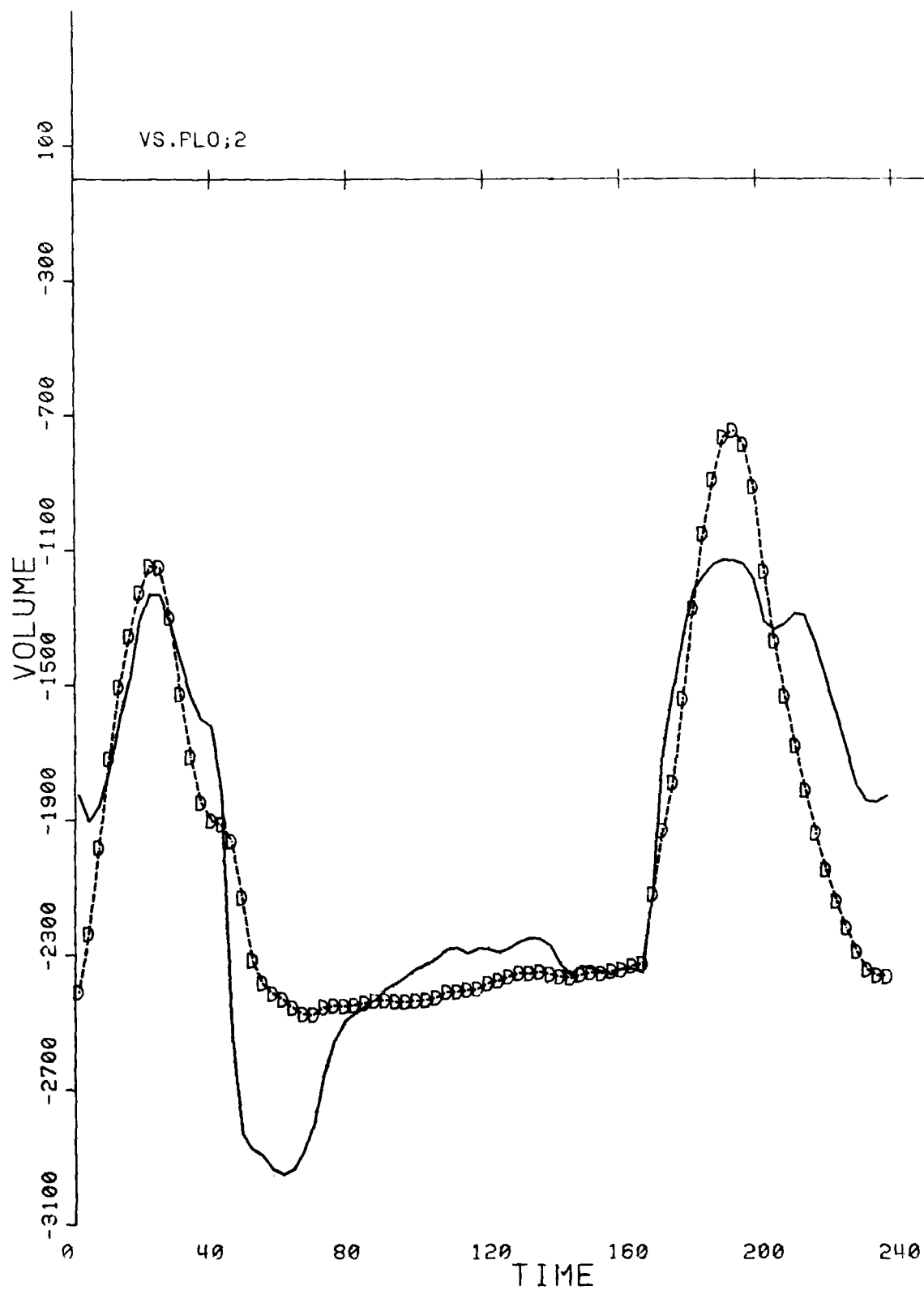


Fig. 14. Valsalva maneuver. Curve fit for Model 2.

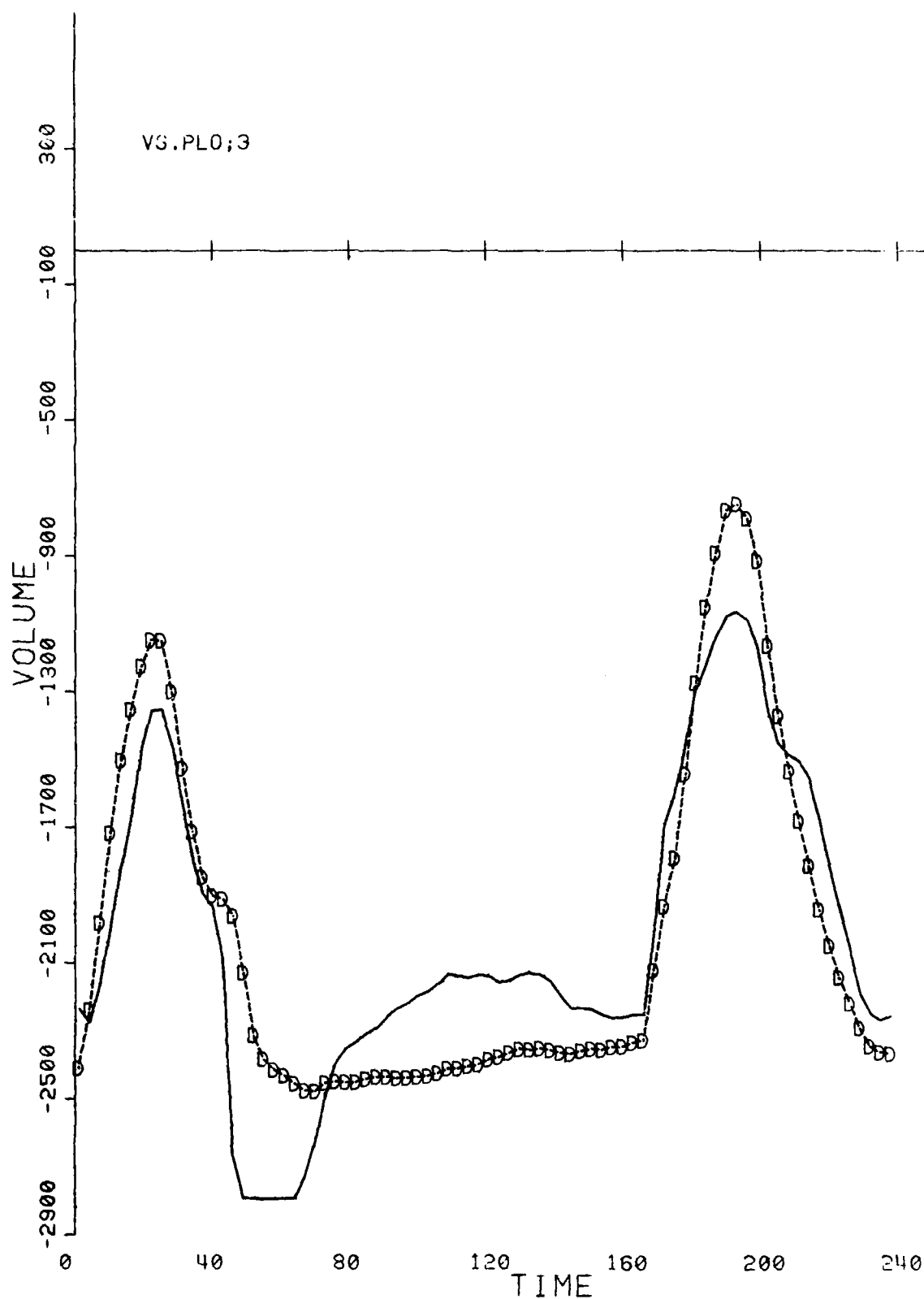


Fig. 15. Valsalva maneuver. Curve fit for Model 3.

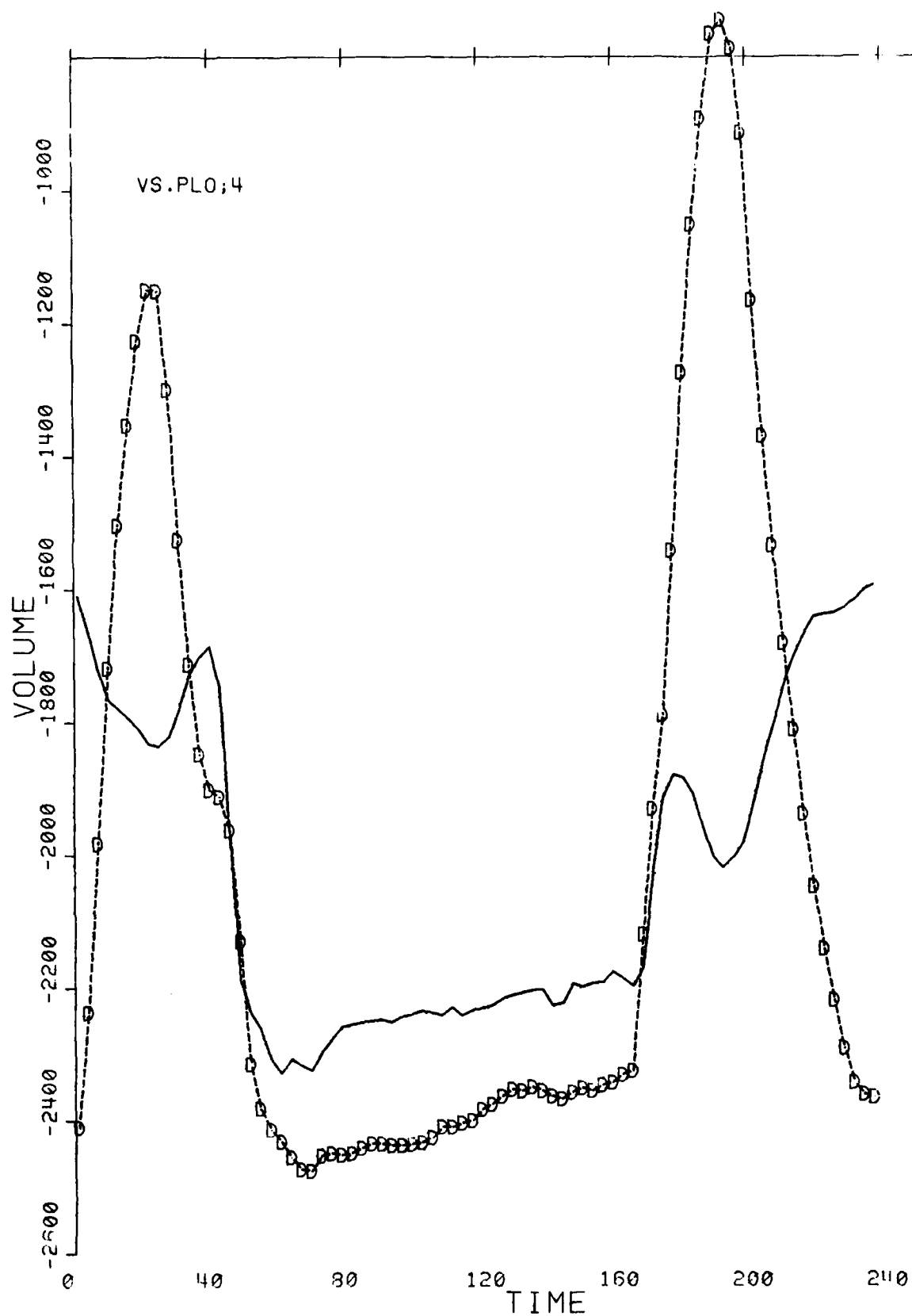


Fig. 16. Valsalva maneuver. Curve fit for Model 4.

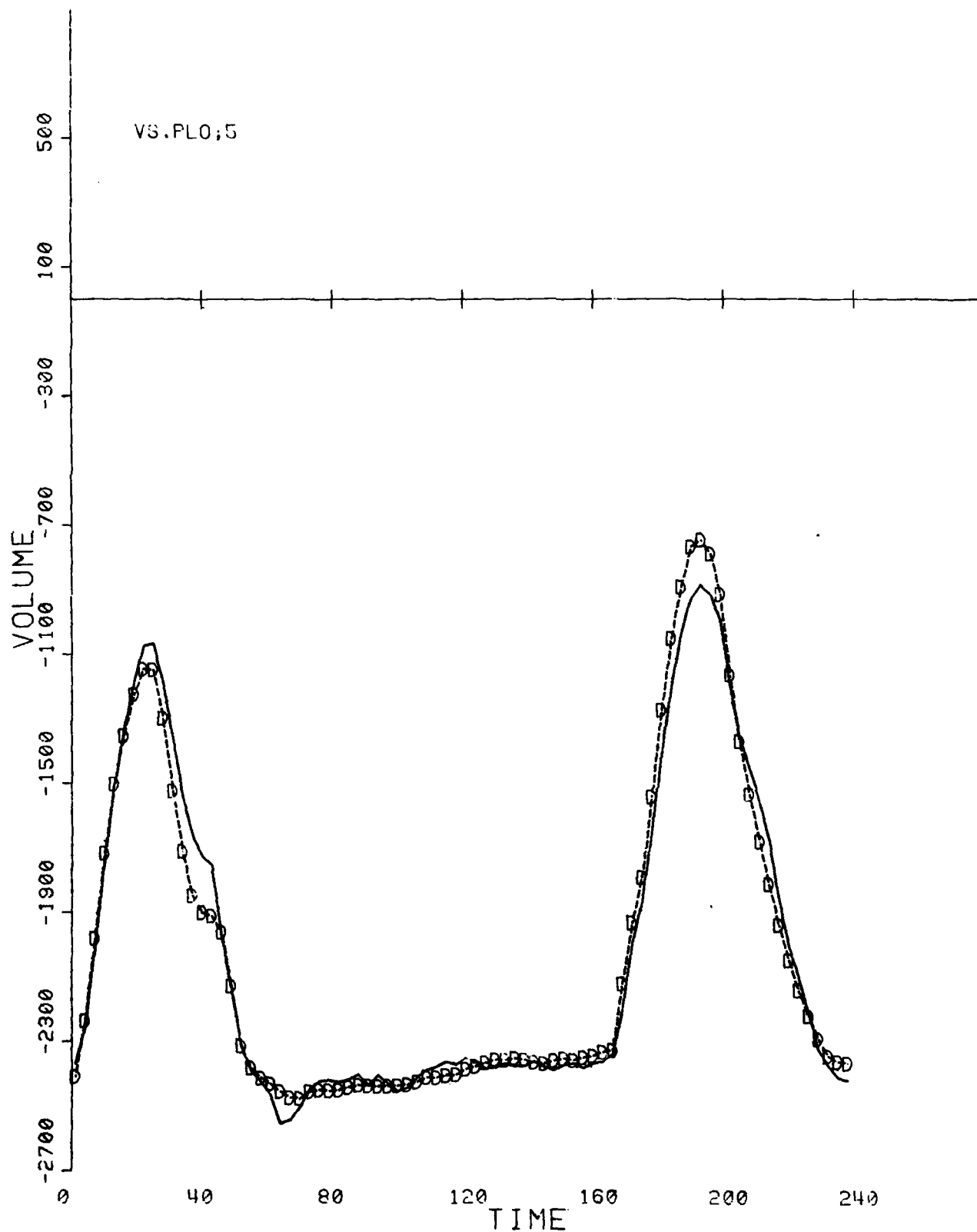


Fig. 17. Valsalva maneuver. Curve fit for Model 5.

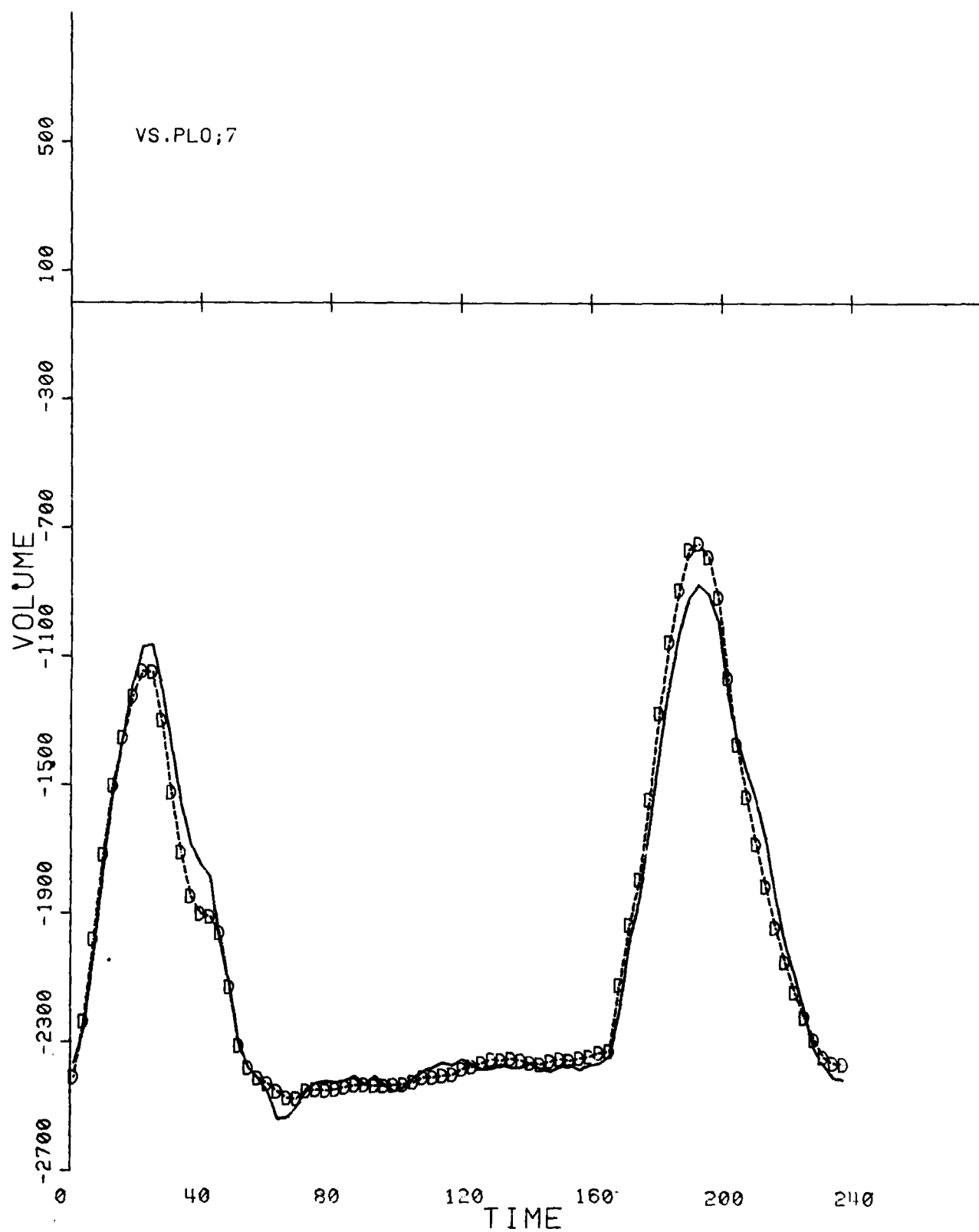


Fig. 18. Valsalva maneuver. Curve fit for Model 7.

TABLE 1. Details of the curve fits for quiet breathing.

MODEL	K_1 (units)	K_2 (units)	K_3 (units)	RMS ERROR (cm^3)
1	31.54 cm	-27.74 cm	---	41.50
2	2487 cm^2	-140.99 cm^2	---	68.00
3	1070 cm^3	-2233 cm^2	---	113.30
4	2606 cm^2	51.14 cm^3	---	74.40
5	55.25 cm^2	5453 cm^3	26040 cm^3	24.70
6	55.25 cm^2	5453 cm^3	139.80 cm^3	24.70
7	44.35 cm	-13.06 cm	32578 cm^3	24.70

to be zero. Shifts in body position or spirometer leakage, although not important in a single breath, may cause considerable discrepancy if unaccounted for in the model. We can eliminate this problem by adding to the data a constant that forces the mean error of the curve fit to be zero.

In Figs. 19 and 20, we present the predictions of Models 5 and 7 on a quiet breath. The constants were derived from a Valsalva maneuver. Notice that the models are essentially equivalent and show good but not excellent agreement.

In Figs. 21 and 22, we see Valsalva maneuvers as predicted from models calibrated with a quiet breath. Notice the curves do not fit very well. It appears, then, that if the magnetometer system is to be used to calculate volumes, it should be "calibrated" with a complex rather than a simple maneuver.

DIMENSIONAL CHANGES DURING NORMAL RESPIRATION

To assess which dimensional changes are important and therefore should be included in future models, we calculated the correlation between changes in magnetometer output and changes in spirometer volume. The first objective was to see whether any parameter had either a very high or very low correlation with the spirometer volume. If so, a model containing only that parameter could be used to predict volume. We analyzed 19 data sets for quiet breathing and 34 data sets for a vital capacity maneuver. We found the magnetometer output and the spirometer output had the following average correlation coefficients. The individual correlations are tabulated in Table 2.

MAGNETOMETER	QUIET	VITAL
A-P CHEST	0.9366	0.9273
LATERAL CHEST	0.6729	0.6857
A-P ABDOMEN	0.9590	0.8528
LATERAL ABDOMEN	0.5027	0.0415

We can see that no single parameter is likely to provide a good estimate of volume. These figures do not imply, however, that any parameter should be omitted. In fact, the necessity for including a parameter can only be determined by comparing models with and without that parameter--with one exception: If the correlation of a parameter with volume is nearly zero for all maneuvers, the parameter can be eliminated.

Another way to reduce the number of parameters in the model without affecting its usefulness is to eliminate a parameter if it is highly correlated with another measured parameter. For example, if A-P chest

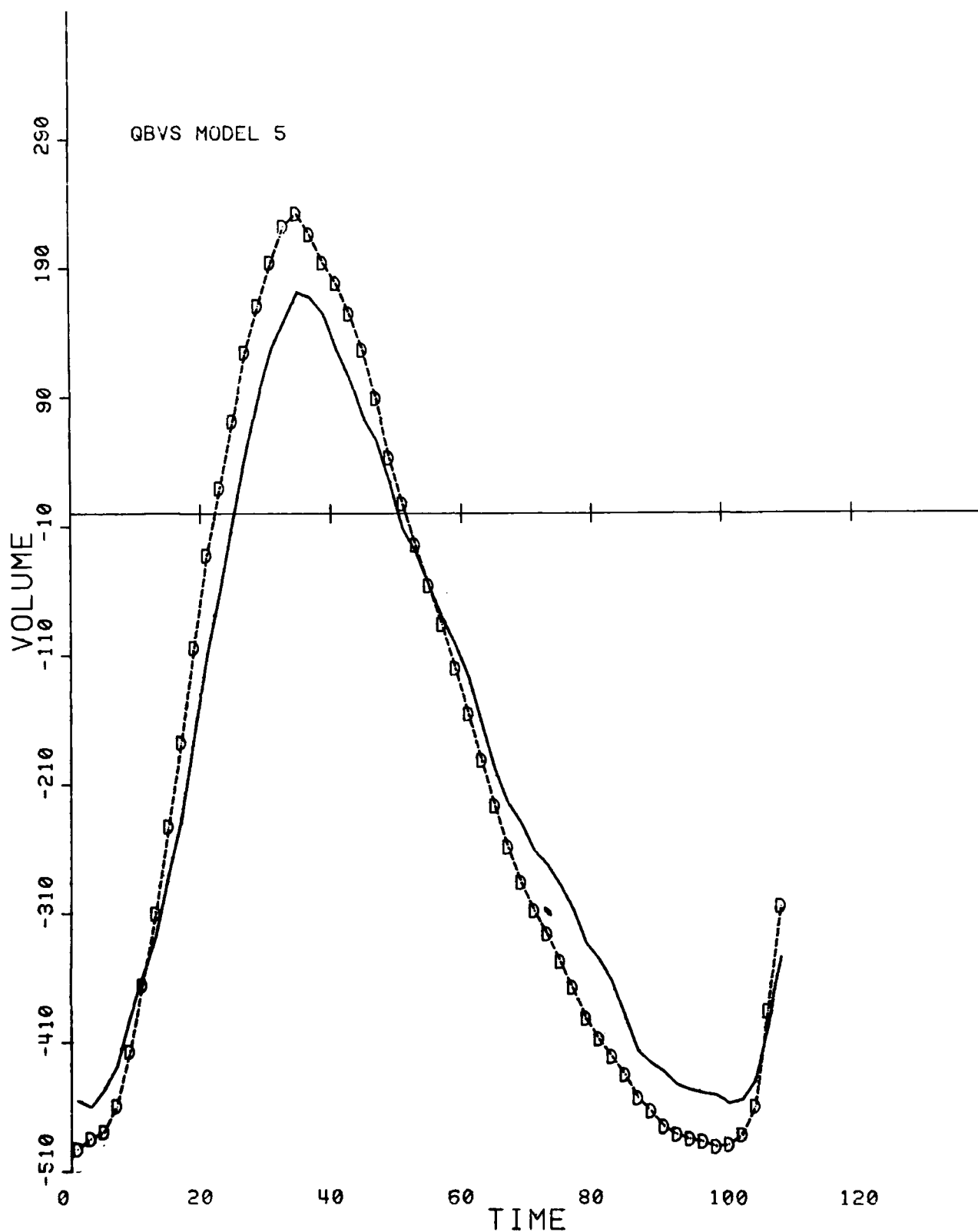


Fig. 19. Model 5. Prediction of quiet breathing using constants calibrated with a Valsalva maneuver. Notice the model underestimates peaks.

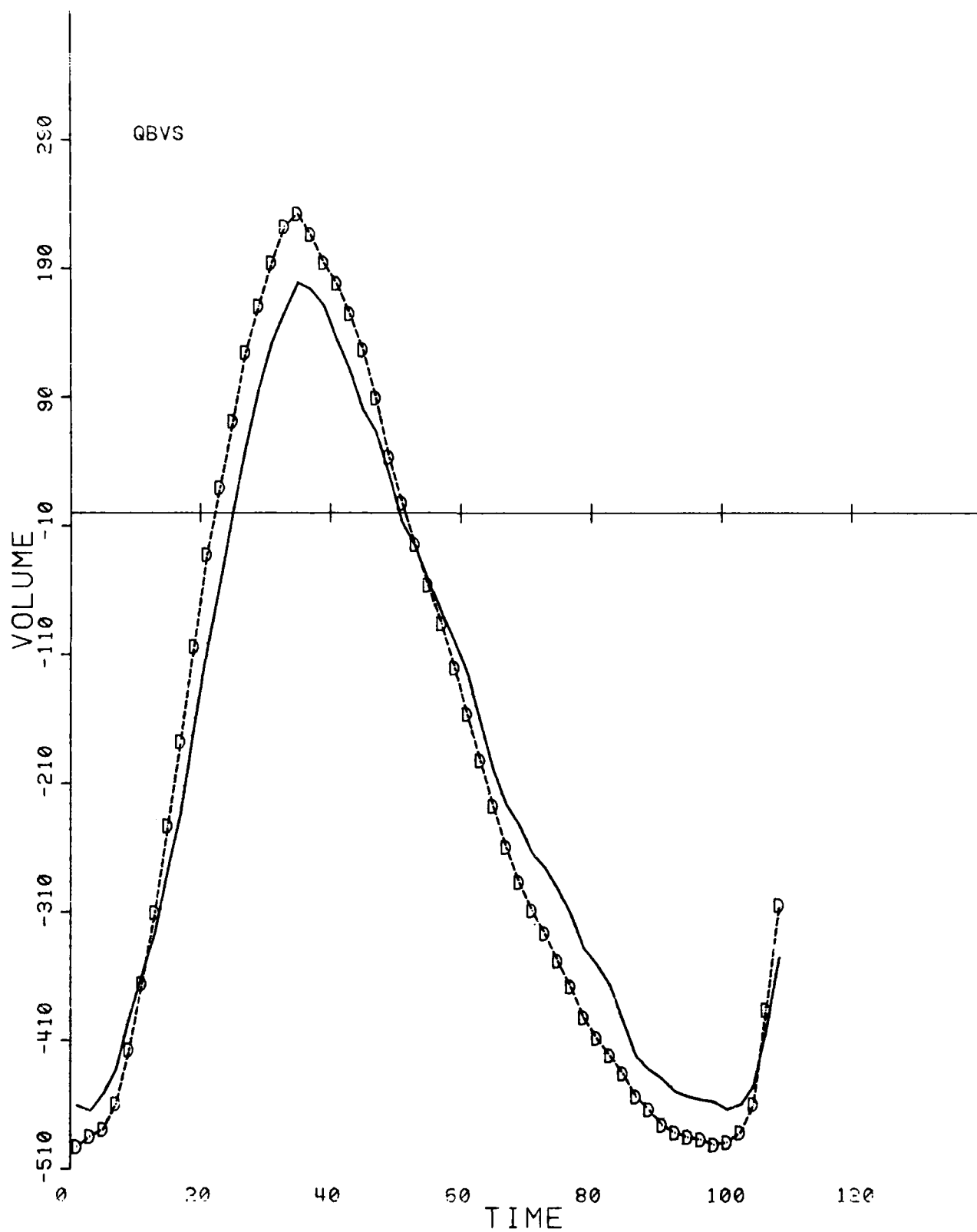


Fig. 20. Model 7. Prediction of quiet breathing using constants calibrated with a Valsalva maneuver. Notice the model again underestimates peaks.

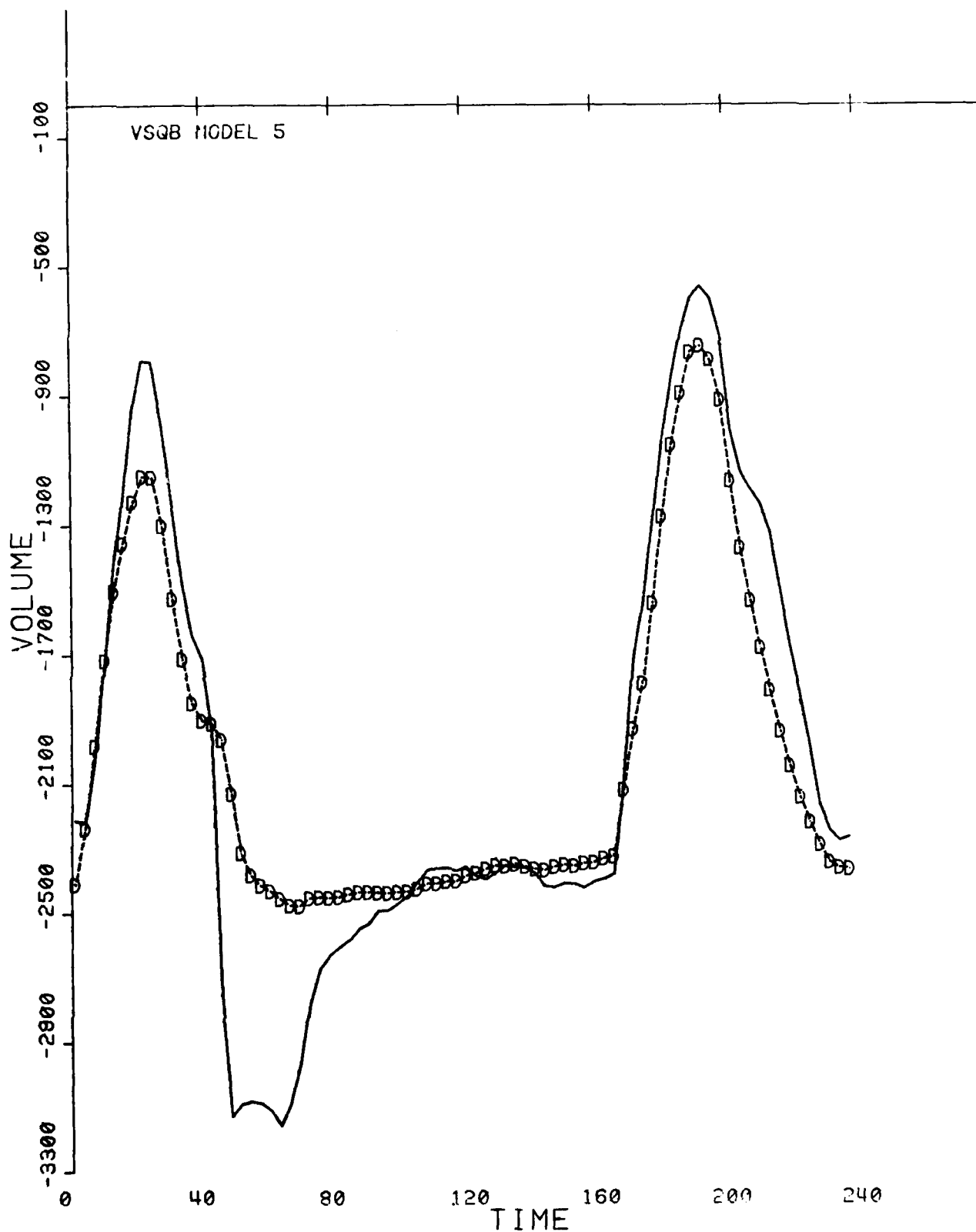


Fig. 21. Model 5. Prediction of Valsalva maneuver using constants calibrated with quiet breathing. In this case the model overestimates peaks.

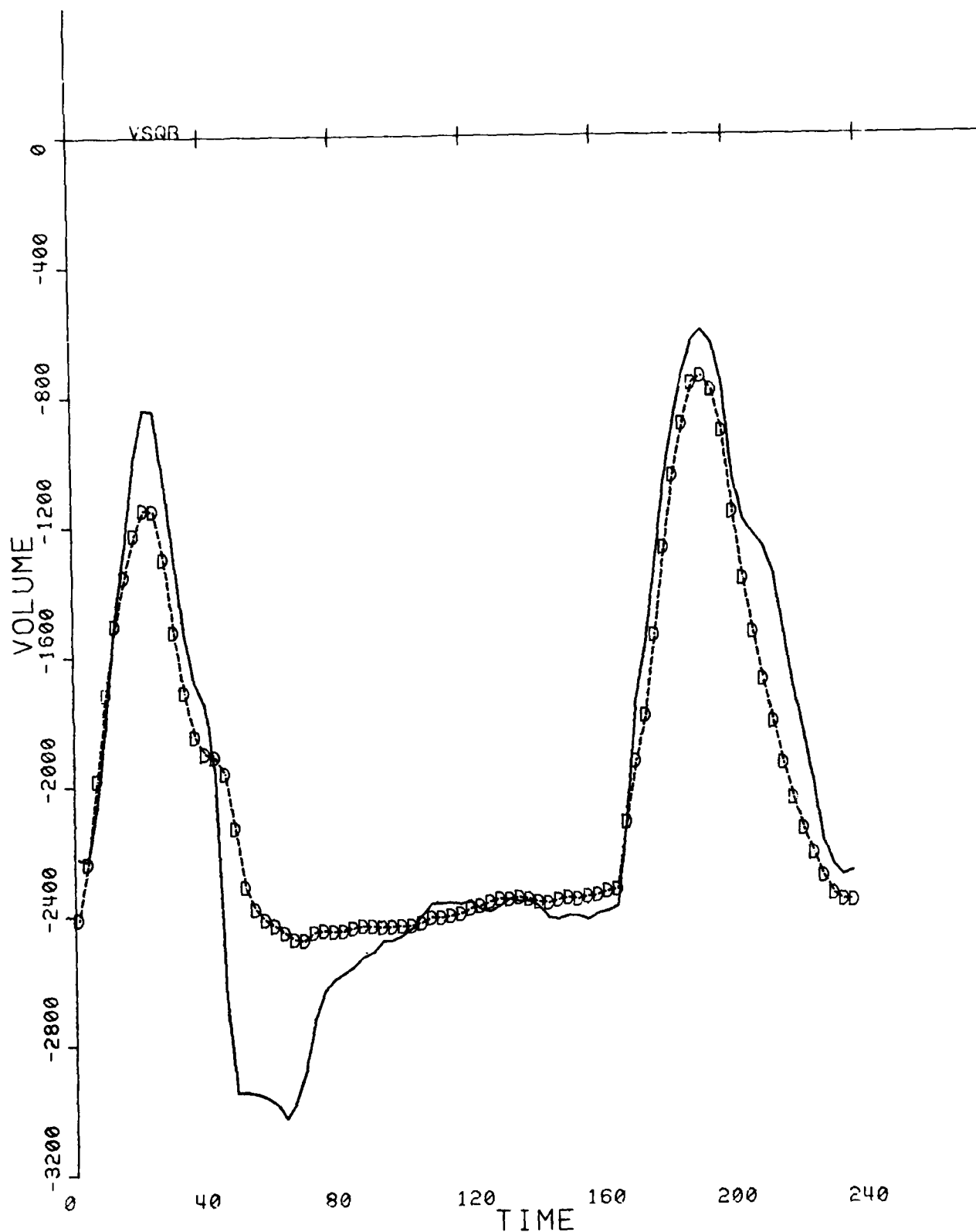


Fig. 22. Model 7. Prediction of Valsalva maneuver using constants calibrated with quiet breathing. Again note the model overestimates peak changes.

Table 2. Correlation Coefficients: Quiet Breathing

SUB	1,2	1,3	1,4	1,5	2,3	2,4	2,5	3,4	3,5	4,5
1	0.9467	0.7120	0.7757	0.9273	0.5493	0.5434	0.8320	0.7757	0.7547	0.7774
2	0.9782	0.9306	0.9652-0.6363	0.8507	0.9068-0.7227	0.9270-0.4474-0.5025				
3	0.7942	0.3245	0.9411	0.1102	0.0425	0.7456-0.0210	0.3091	0.4191-0.0949		
4	0.9573	0.5831	0.9698-0.2813	0.4204	0.8835-0.4323	0.5960	0.0580-0.2651			
5	0.9395	0.6930	0.8791-0.4524	0.6088	0.7359-0.4854	0.6218-0.3216-0.1487				
6	0.9394	0.6929	0.8799-0.4527	0.6087	0.7366-0.4858	0.6220-0.3224-0.1521				
7	0.9649	0.9178	0.9443	0.3291	0.8870	0.8670	0.1342	0.8009	0.2698	0.5073
8	0.9572	0.9151	0.8969	0.3557	0.8394	0.8115	0.1255	0.7676	0.4192	0.4520
9	0.9754	0.9796	0.7532	0.8431	0.9700	0.6191	0.7781	0.7114	0.8712	0.7355
10	0.9928	0.9896	0.7990	0.9004	0.9817	0.7607	0.9032	0.7879	0.9163	0.5985
11	0.9749-0.0107	0.8817-0.4940-0.1436	0.7800-0.6372	0.1179	0.5231-0.2009					
12	0.9546	0.1151	0.8498-0.5210-0.1228	0.6732-0.6697	0.4007	0.3705-0.2118				
13	0.9609	0.9170	0.8636-0.6666	0.9156	0.7737-0.7030	0.7342-0.6170-0.3531				
14	0.9587	0.9170	0.8742-0.6633	0.9142	0.7835-0.6994	0.7451-0.6111-0.3662				
15	0.9456	0.7107	0.9148	0.7456	0.7030	0.7392	0.5519	0.6353	0.6823	0.8882
16	0.9431	0.6918	0.8488	0.4099	0.6063	0.6398	0.1018	0.6930	0.5035	0.7710
17	0.9976-0.3269	0.8775-0.6882-0.3520	0.8631-0.6943	0.0343	0.7132-0.4439					
18	0.9959-0.1432	0.9062-0.5153-0.2091	0.8719-0.5690	0.2575	0.6376-0.2080					
19	0.9611	0.9591	0.5057	0.1002	0.9008	0.2910-0.1365	0.4865	0.0930	0.8673	
20	0.9831	0.9673	0.7528	0.2283	0.9335	0.6788	0.1096	0.6526	0.1978	0.6505
21	0.9228	0.8942	0.7641	0.0407	0.7404	0.4893-0.1170	0.8154	0.0796	0.2749	
22	0.9209-0.8323	0.6967	0.1201	0.7152	0.4475	0.0016	0.6404	0.1743	0.2568	
23	0.8911	0.4780	0.8474	0.3002	0.0811	0.5326-0.0695	0.7543	0.8244	0.5242	
24	0.9150	0.5787	0.8512	0.4638	0.2449	0.5814	0.1757	0.8067	0.8884	0.6244
25	0.8898	0.4407	0.8104	0.2800	0.0360	0.4692-0.0662	0.7934	0.7347	0.5321	
26	0.9021	0.9156	0.7053	0.5898	0.7765	0.3826	0.3830	0.6456	0.7038	0.6213

Table 2. Correlation Coefficients: Quiet Breathing (continued)

27	0.8814	0.8720	0.7436	0.8001	0.7438	0.3842	0.6251	0.6408	0.8333	0.7451
28	0.9110	0.8348	0.9357-0.8926	0.8013	0.7205-0.9490	0.7050-0.6811-0.7678				
29	0.9098	0.8975	0.8994-0.9135	0.9414	0.6459-0.9818	0.6851-0.9426-0.6876				
30	0.9800	0.8008	0.9115-0.6981	0.7022	0.8242-0.8157	0.8400-0.2674-0.3705				
31	0.9593	0.8104	0.8999-0.5894	0.7220	0.7653-0.7633	0.7190-0.3977-0.2377				
32	0.3752	0.8940	0.9477	0.7167	0.3584	0.5165	0.4032	0.8612	0.8568	0.8191
33	0.9786	0.5884	0.9575	0.8490	0.6563	0.9265	0.8129	0.4110	0.4422	0.8125
34	0.9720	0.9403	0.9459	0.7670	0.9047	0.8804	0.6364	0.8701	0.7207	0.8517

AVERAGE CORRELATION

34	0.9273	0.6857	0.8528	0.0415	0.5685	0.6844-0.1013	0.6431	0.2670	0.2147
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Table 2. Correlation Coefficients: Vital Capacity

SUB	1,2	1,3	1,4	1,5	2,3	2,4	2,5	3,4	3,5	4,5
1	0.9441	0.8617	0.9366	0.8193	0.8022	0.9186	0.6252	0.8946	0.7480	0.6850
2	0.9549	0.9698	0.9852	0.9833	0.9490	0.9477	0.9230	0.9385	0.9672	0.9590
3	0.8958	0.9466	0.9713	-0.8425	0.8565	0.8375	-0.8081	0.9160	-0.7864	-0.8054
4	0.9386	0.9446	0.9493	0.7356	0.8530	0.9814	0.7417	0.8461	0.5664	0.8221
5	0.9606	0.8260	0.9189	0.9225	0.7363	0.8651	0.8702	0.9413	0.9397	0.9612
6	0.9775	0.8825	0.9602	0.9276	0.8967	0.9783	0.9302	0.9099	0.9587	0.9388
7	0.9931	0.7703	0.9928	0.0107	0.8098	0.9812	-0.0313	0.7042	-0.3829	0.0790
8	0.7924	0.8988	0.9980	-0.6239	0.8866	0.9915	-0.6681	0.8975	-0.5435	-0.6294
9	0.9834	-0.1779	0.9941	0.9746	-0.2251	0.9684	0.9314	-0.1139	-0.0832	0.9873
10	0.9900	-0.8577	0.9980	0.9835	-0.8535	0.9842	0.9590	-0.8502	-0.8372	0.9897
11	0.9692	0.6392	0.8871	0.6904	0.5801	0.8744	0.6315	0.4805	0.3979	0.7826
12	0.9671	0.9583	0.9860	0.8838	0.9155	0.9602	0.8554	0.9703	0.9137	0.8983
13	0.7459	0.9148	0.9586	-0.2239	0.9276	0.8643	-0.1424	0.8039	-0.1745	-0.2070
14	0.7632	0.9489	0.9598	0.7762	0.8849	0.8819	0.7442	0.9089	0.8533	0.6729
15	0.7335	0.7828	0.9862	0.9549	0.9015	0.8979	0.8062	0.9760	0.9685	0.9618
16	0.7604	-0.4588	0.8998	-0.8525	-0.4086	0.9116	-0.7945	-0.4780	0.4819	-0.8472
17	0.6524	0.9032	0.9864	0.9029	0.6289	0.5356	0.6475	0.8742	0.7214	0.8691
18	0.7666	0.9563	0.8813	0.8674	0.7402	0.8286	0.8150	0.8409	0.8535	0.9530
19	0.9864	0.8758	0.9719	0.6610	0.8817	0.9407	0.6441	0.7794	0.8164	0.6087

AVERAGE CORRELATION

19	0.9366	0.6729	0.9590	0.5027	0.6191	0.9026	0.4568	0.6442	0.3884	0.5095
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changes and A-P abdomen changes are highly correlated, then only one of them is necessary in the model. The correlations between the magnetometer pairs for the data sets described previously are summarized below:

MAGNETOMETER PAIR	QUIET	VITAL
A-P CHEST--LATERAL CHEST	0.6191	0.5685
A-P CHEST--A-P ABDOMEN	0.9026	0.6844
A-P CHEST--LATERAL ABDOMEN	0.4568	-0.1013
LATERAL CHEST--A-P ABDOMEN	0.6442	0.6431
LATERAL CHEST--LATERAL ABDOMEN	0.3884	0.2670
A-P ABDOMEN--LATERAL ABDOMEN	0.5095	0.2147

It appears from the above table that for quiet breathing we may be able to eliminate either the A-P chest magnetometer or the A-P abdomen magnetometer and still have a useful model. If, however, we want the model to be useful for more complicated maneuvers, it would not be wise to eliminate either magnetometer.

To visualize the dependence of volume on changes in dimensions and to better assess the correlation between dimensional changes, refer to Figs. 23 through 32, where several crossplots of typical data are presented. Notice that in almost all cases, hysteresis is present showing differences between expiration and inspiration. Given this hysteresis, it is remarkable that the simple models used in this study fit the data so well. In Fig. 33, a typical output versus time curve is plotted for tidal breathing.

The complexity of chest and abdomen movement during respiration is apparent in Figs. 34 and 35. Fig. 34 plots the change in the A-P diameters during a Valsalva maneuver. In Fig. 35, the change in lateral diameter of the abdomen during a vital capacity maneuver is plotted. In view of these complex changes, it is not surprising that the simple two parameter models fail to predict the volume accurately even during a calibration run. In fact, it is quite surprising that the three parameter models do so well.

CONCLUSIONS

Based on the present study we can make the following conclusions:

- 1) The magnetometer system is relatively insensitive to angular changes except about the axis joining the centers of the pair. Rotation about this axis must be avoided.
- 2) A linear analysis is possible to determine the unknown constants in the volume models.

- 3) Any two or three parameter model is adequate to predict volume for the breath from which the constants are determined.
- 4) Three parameter models are necessary for predictive rather than calibrating models.
- 5) Calibration should be done with a complex maneuver such as the Valsalva if the model is required to predict complicated maneuvers.
- 6) Sets of model constants have a unique meaning but individual constants do not.
- 7) No dimensional change is highly enough correlated with volume changes to be used alone.
- 8) No two dimensional changes are highly enough correlated to eliminate either from the model.
- 9) Hysteresis of respiratory movement is apparent and should be considered in any future modeling studies.

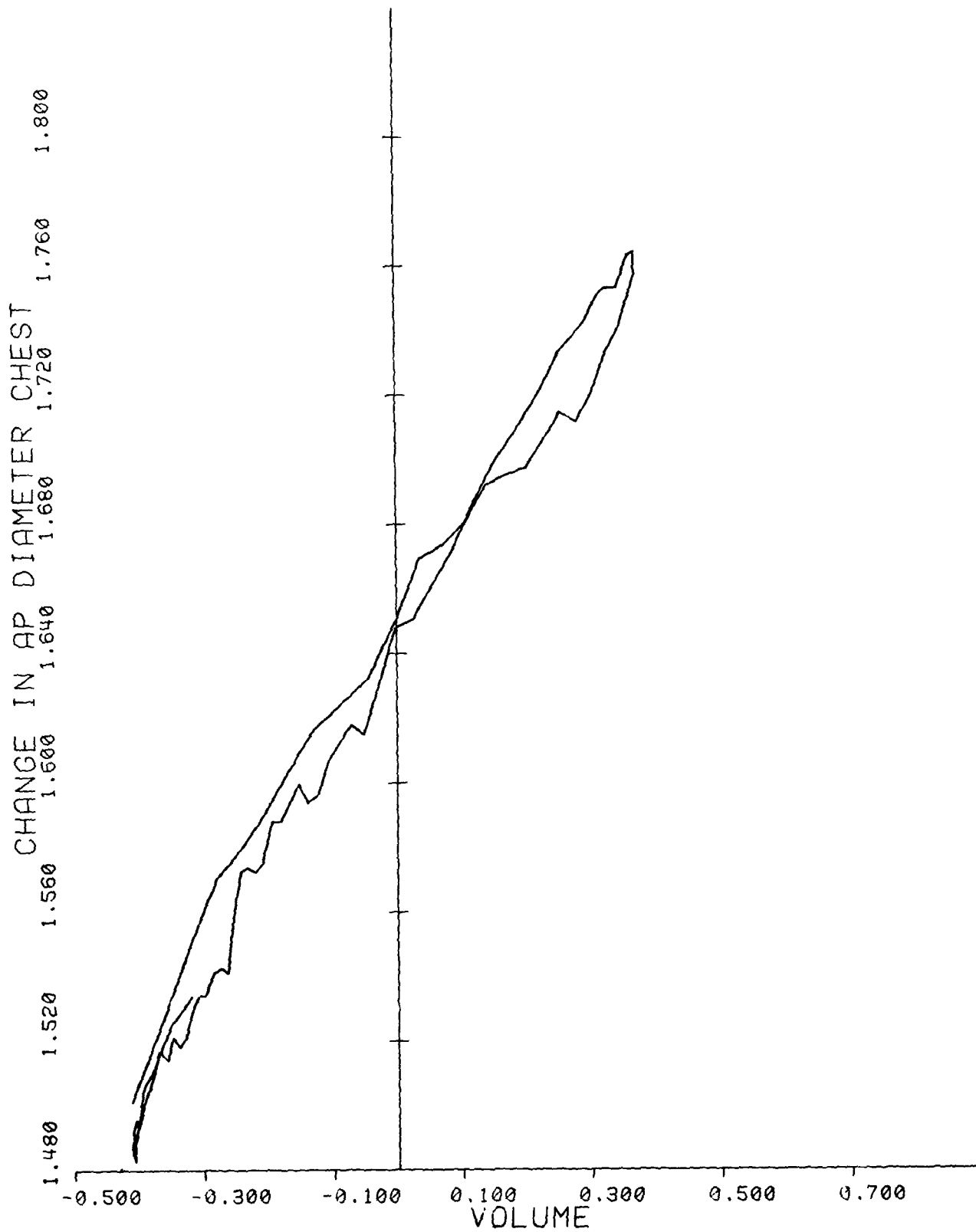


Fig. 23. Change in the A-P diameter of the chest versus volume for tidal breathing. Notice there is good correlation and only slight hysteresis. Curves 23 through 32 are tidal breathing.

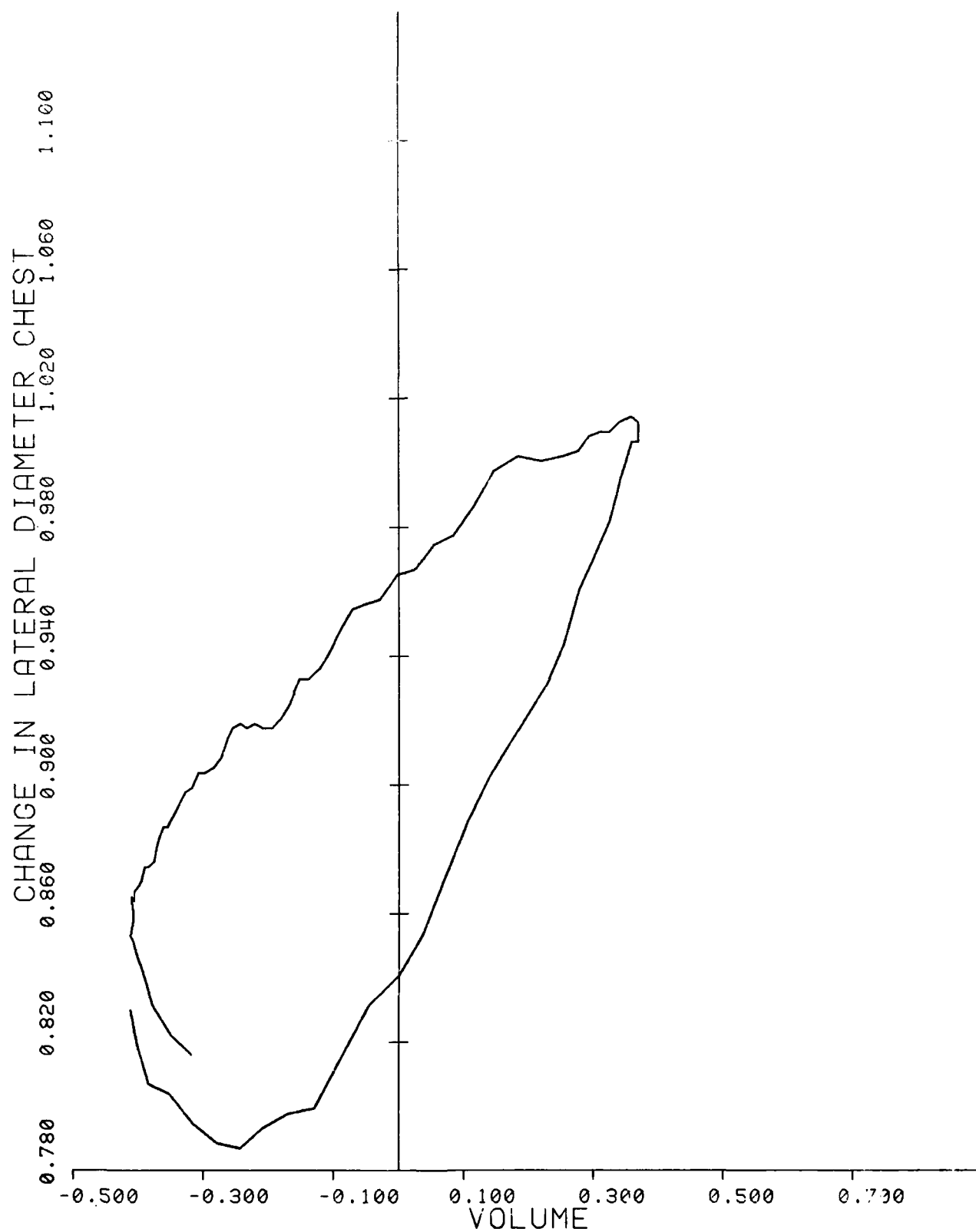


Fig. 24. Change in the lateral diameter of the chest as a function of volume. Notice the large hysteresis.

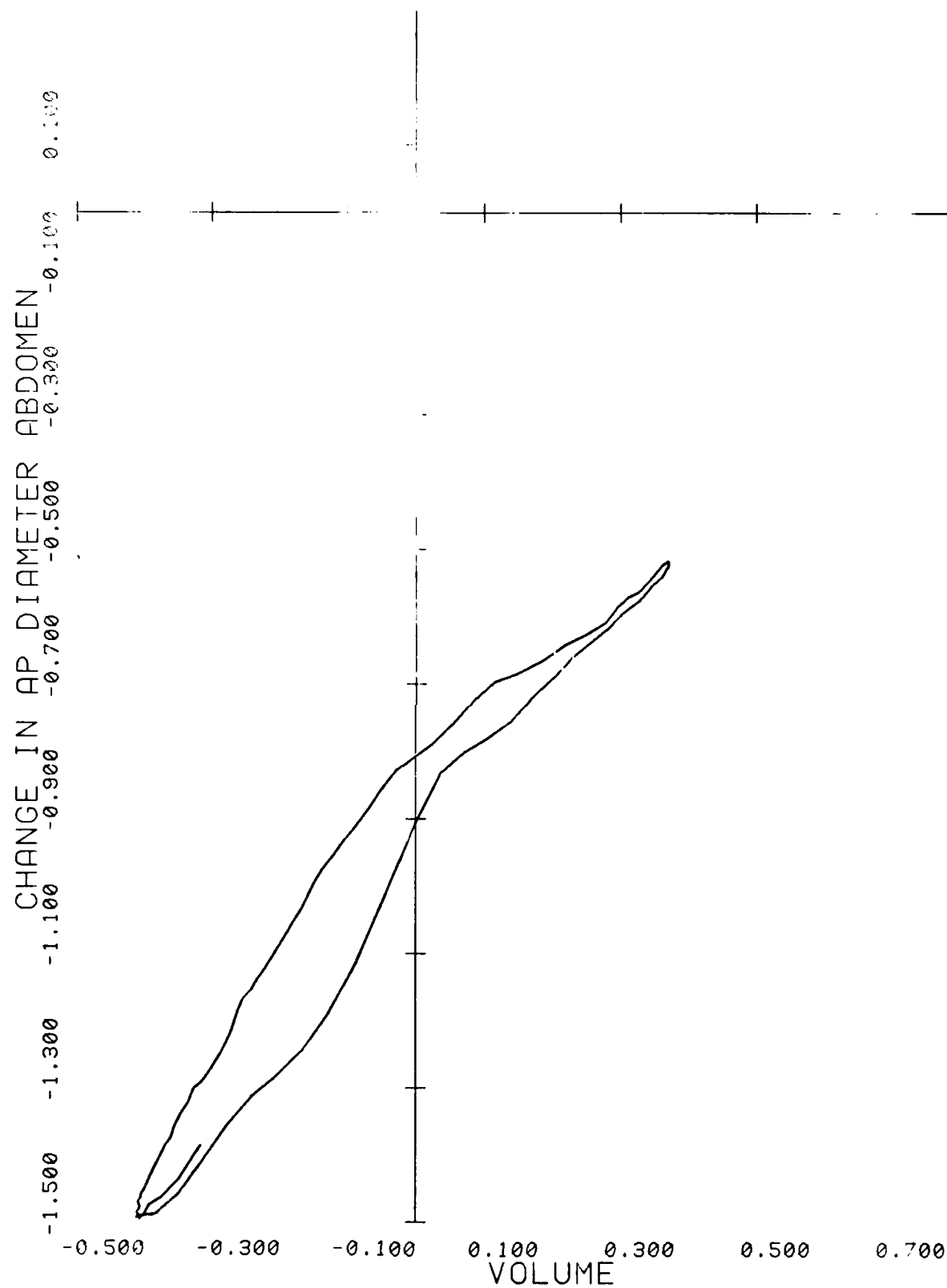


Fig. 25. Change in the A-P diameter of the abdomen with volume.

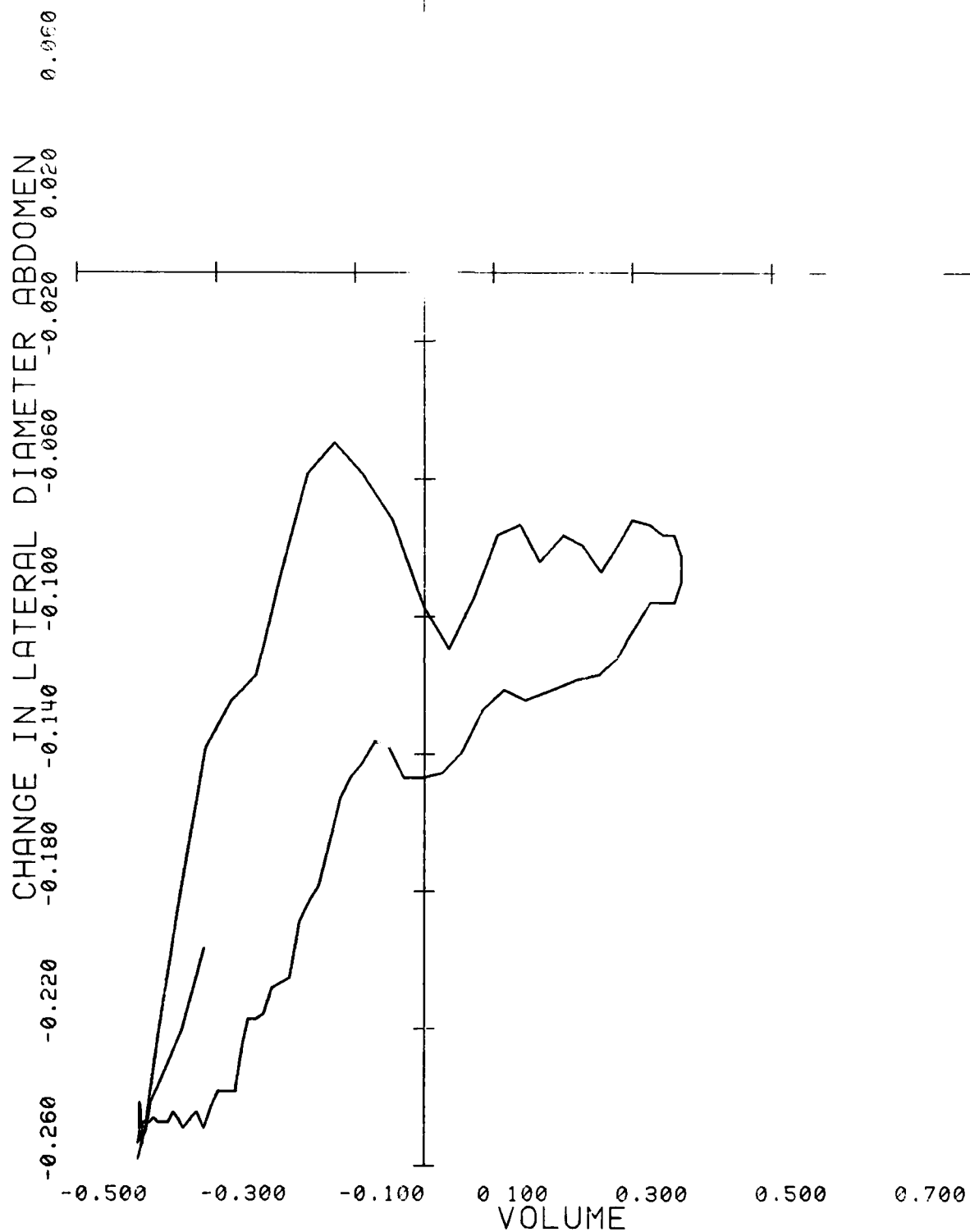


Fig. 26. Change in the lateral diameter of the abdomen with volume. Notice the diameter increases at low volumes and then holds rather constant at high volumes.

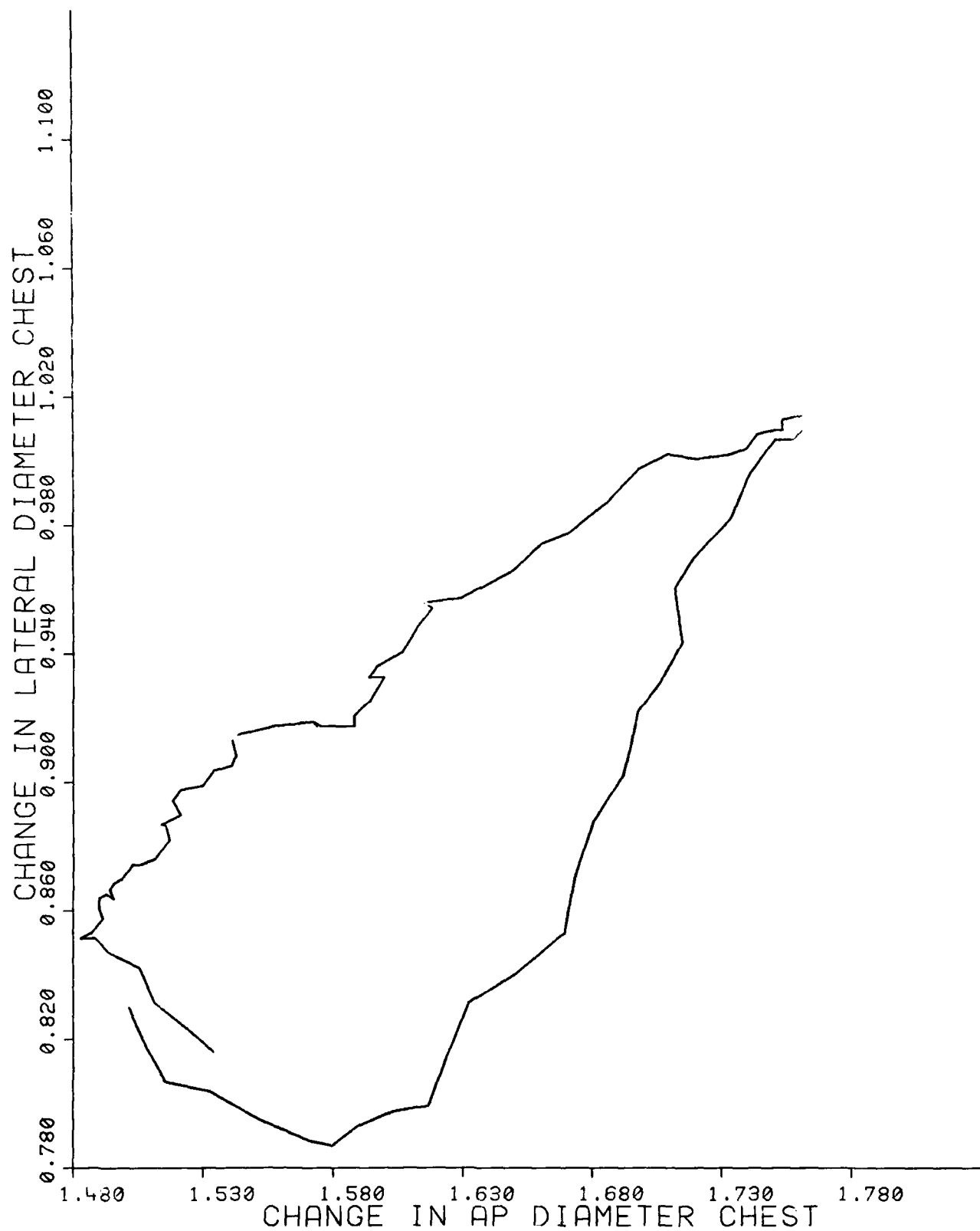


Fig. 27. Change in the lateral diameter of the chest with changing A-P diameter of the chest. Notice there is a large amount of hysteresis showing that it is unlikely that a lag in the spirometer causes the hysteresis in the earlier figures.

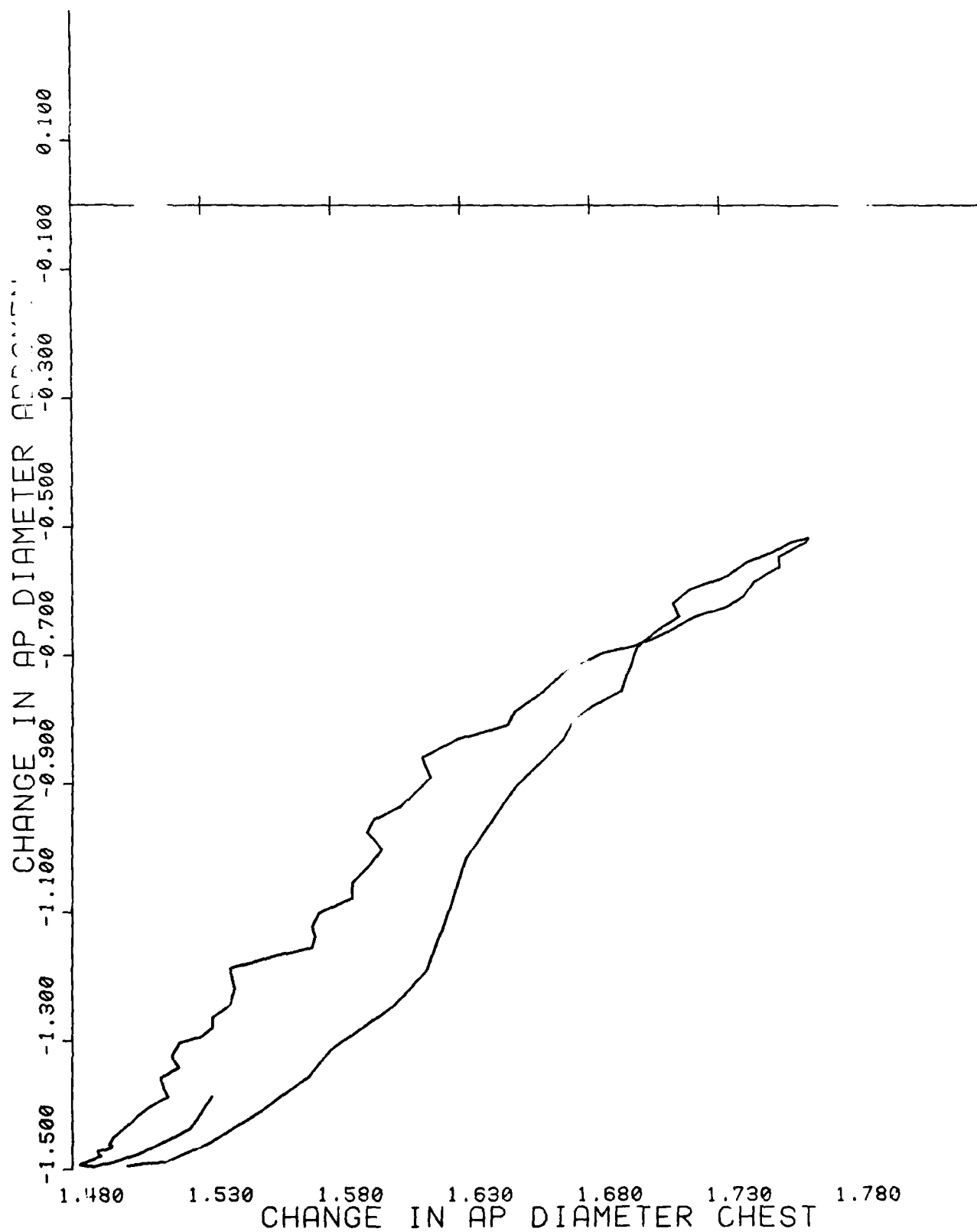


Fig. 28. Change in A-P diameter of the abdomen as the A-P diameter of the chest changes. Notices the relatively high correlation.

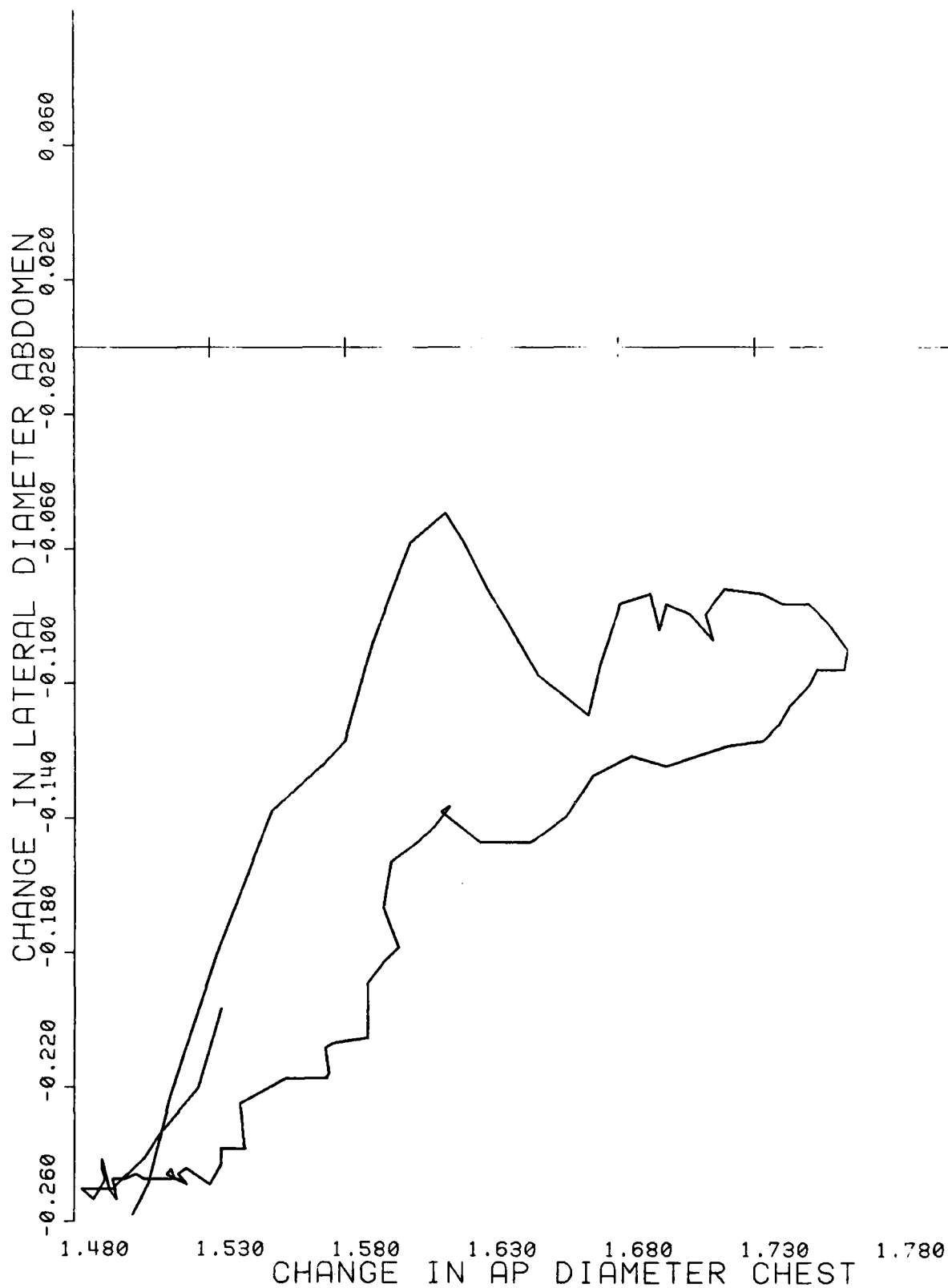


Fig. 29. Change in the lateral diameter of the abdomen with change in the A-P diameter of the chest showing large hysteresis and low correlation.

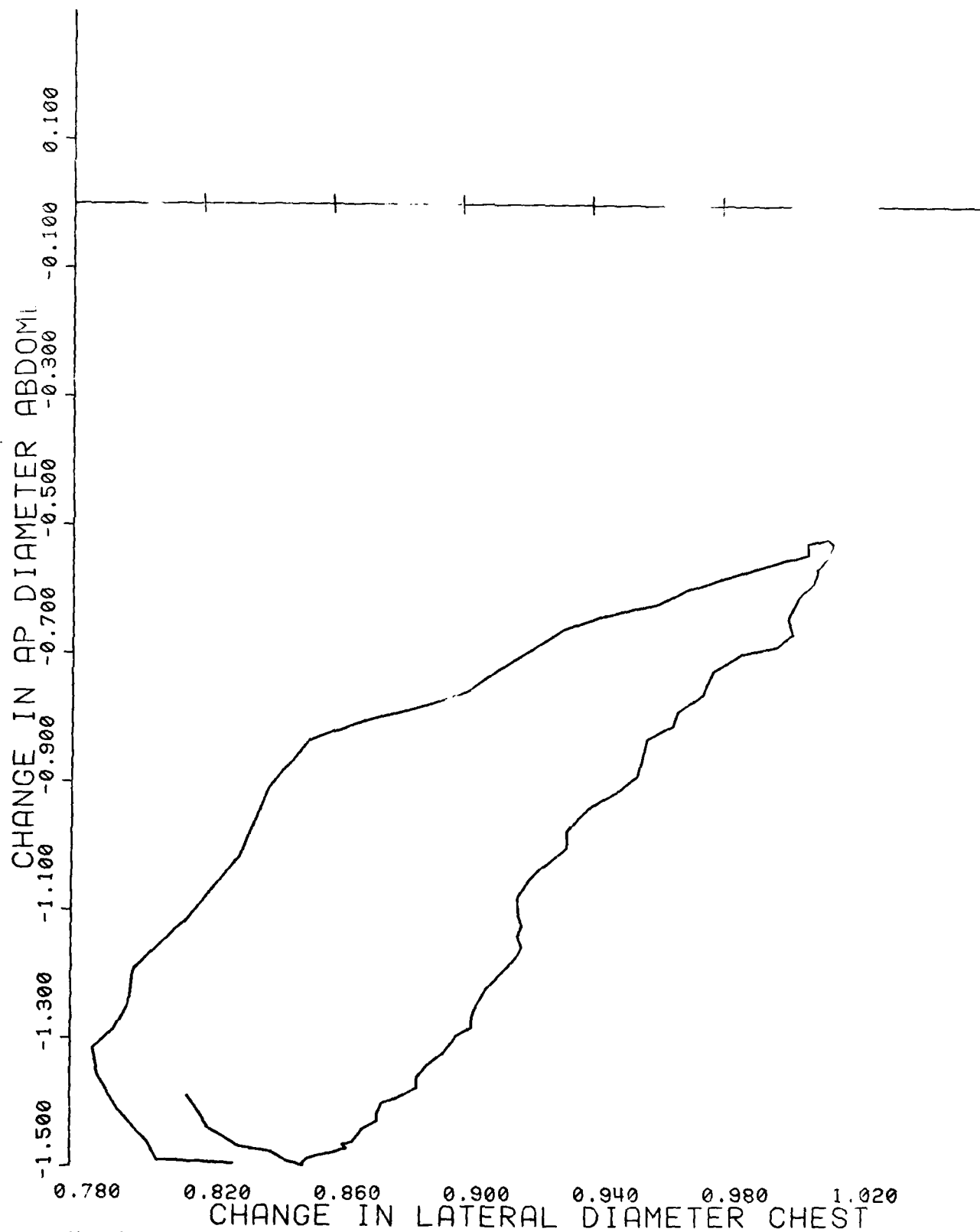


Fig. 30. Change in the A-P diameter of the abdomen with changes in the lateral diameter of the chest.

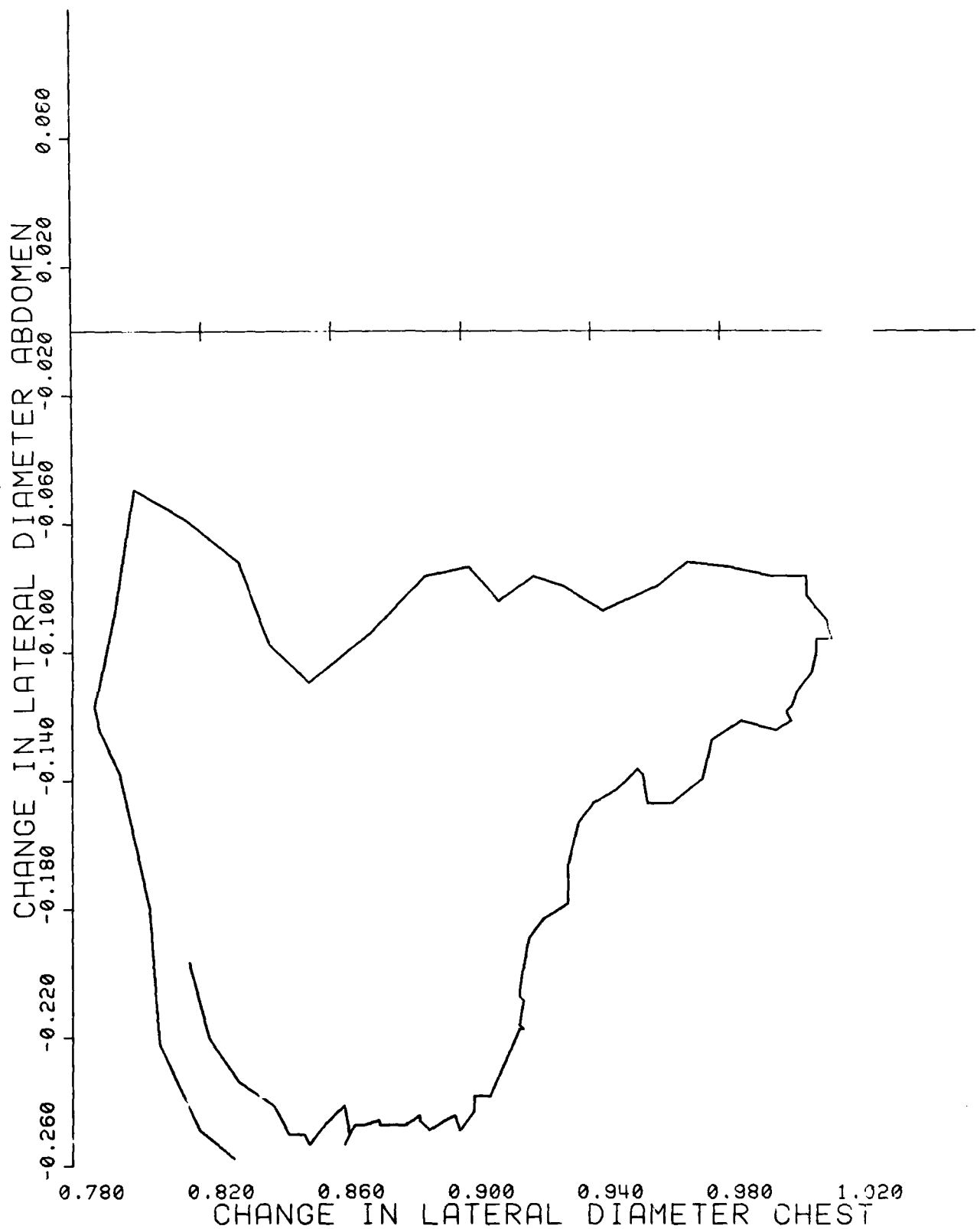


Fig. 31. Change in the lateral diameter of the abdomen with change in the lateral diameter of the chest. Notice the difference in trend between inspiration and expiration.

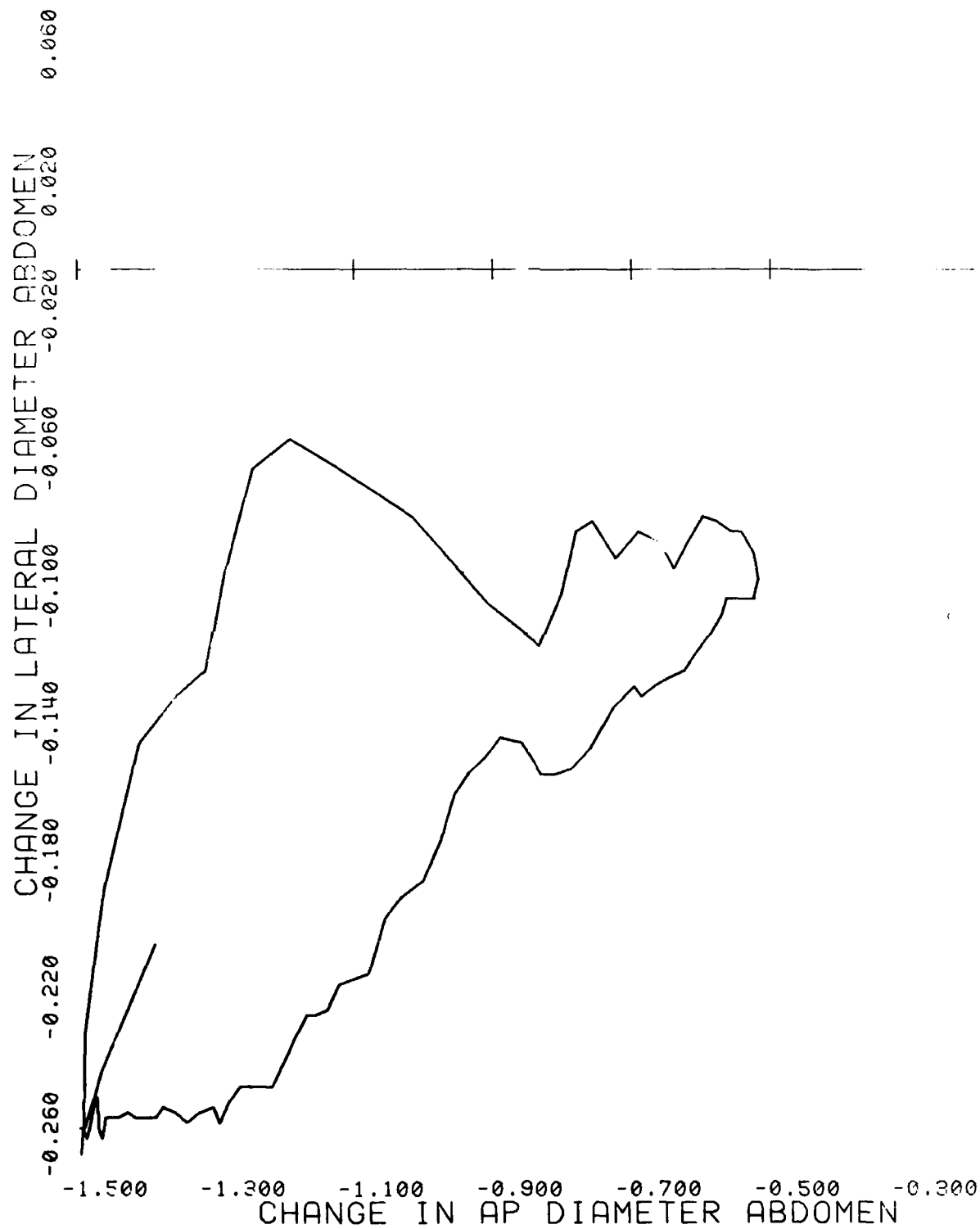


Fig. 32. Change in the lateral diameter of the abdomen with changing A-P diameter of the abdomen.

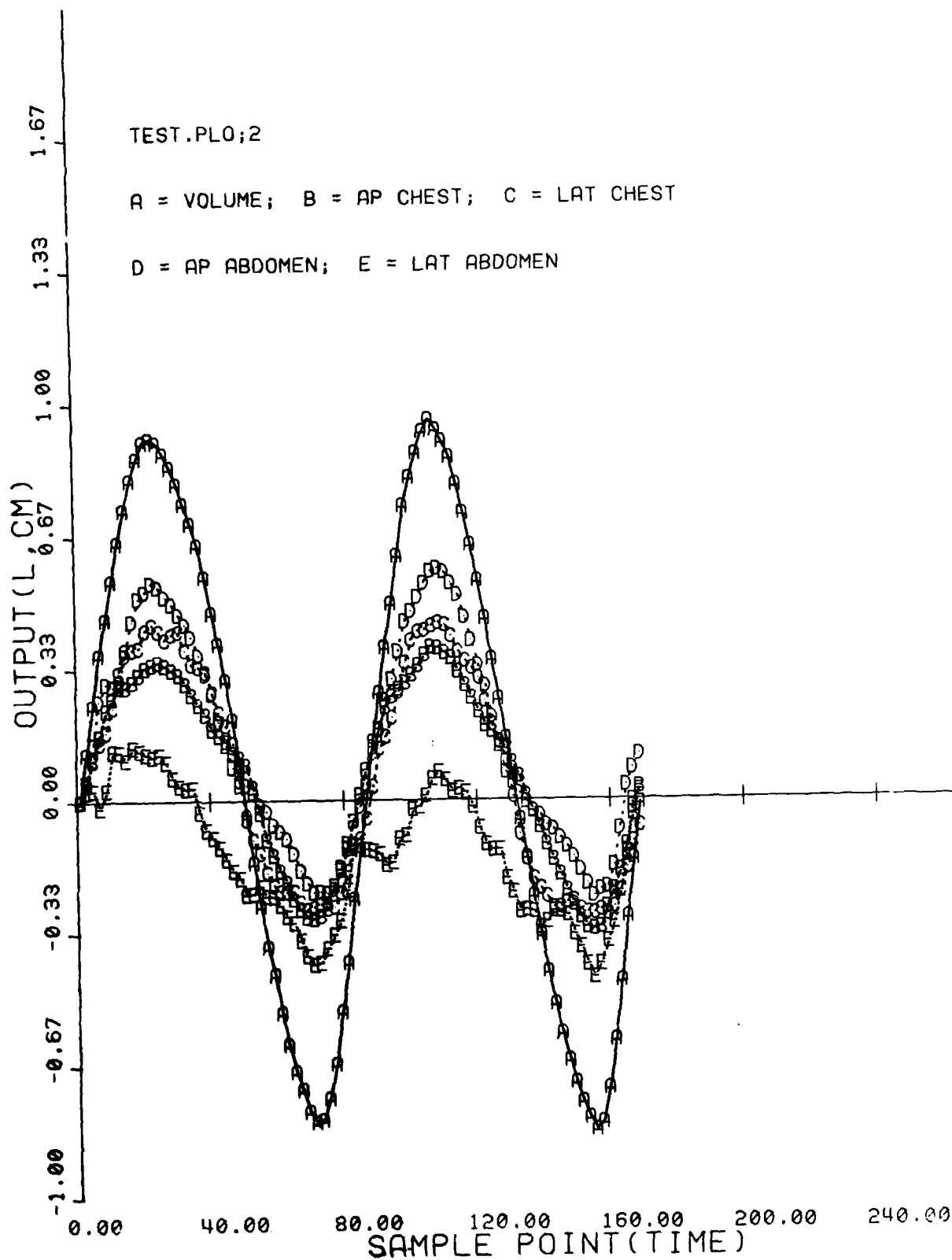


Fig. 33. Typical tidal breathing maneuver. Magnetometers "B" and "C" are at the nipple level. Magnetometers "D" and "E" are at the level of the umbilicus.

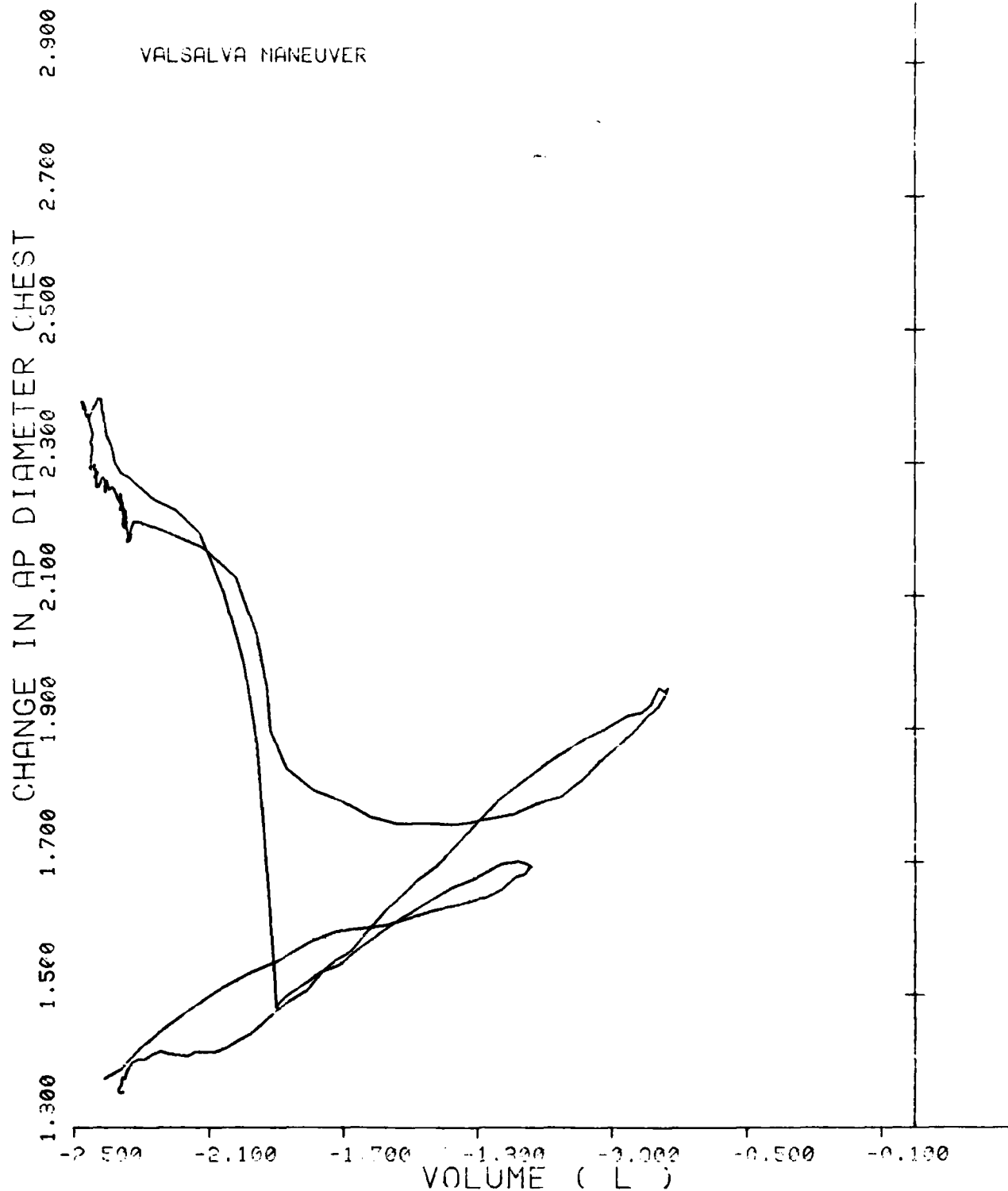


Fig. 34. Change in the A-P diameter of the chest with volume during a Valsalva maneuver showing complex changes. Compare this with the simple curve for tidal breathing (Fig. 23).

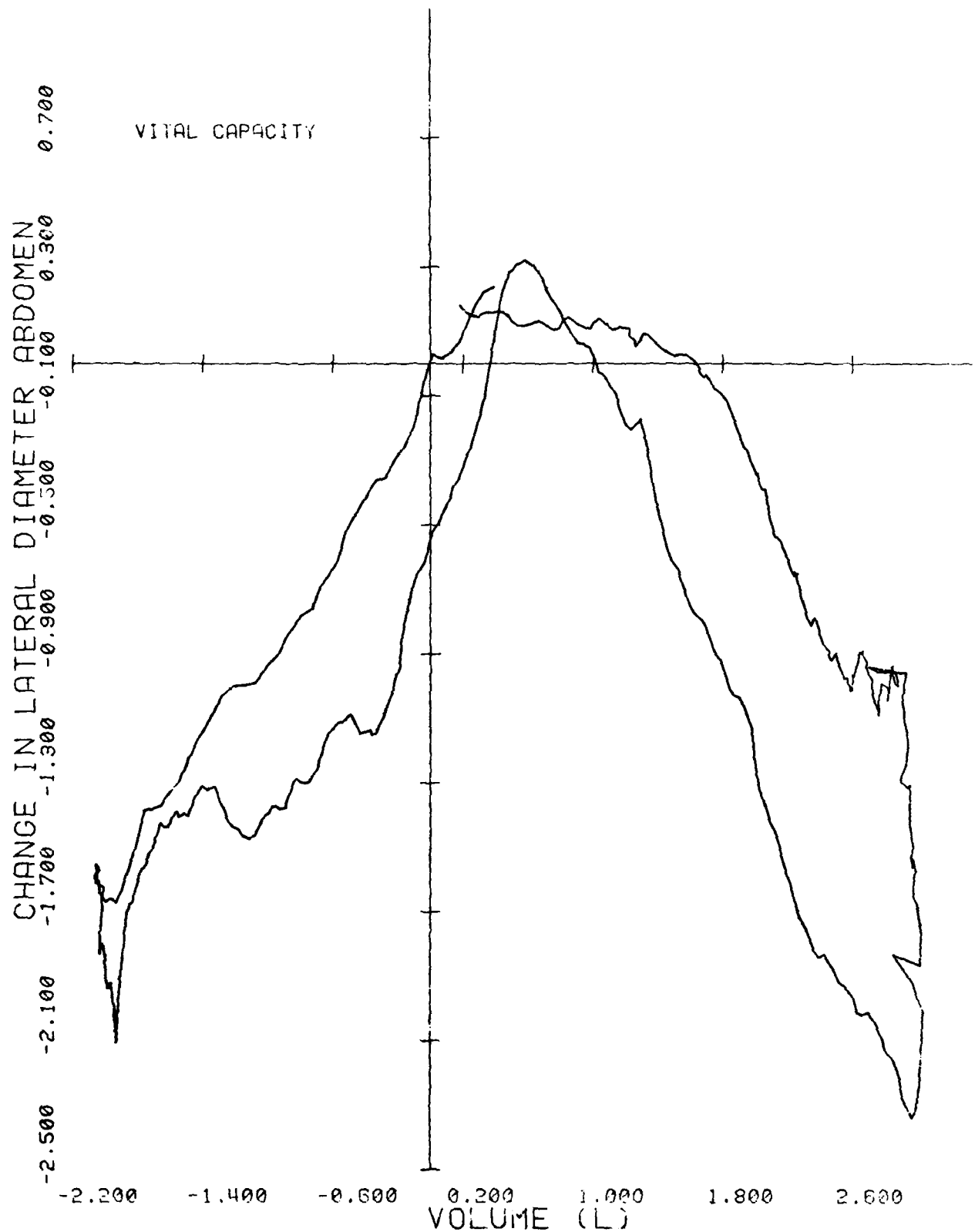


Fig. 35. Change in lateral diameter of the abdomen with volume for a vital capacity maneuver. Notice the abdomen first increases in size laterally as volume increases and then decreases with further increases in volume.

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